HOWARD, R. LEE., Ph.D. Kinematic and Kinetic Effects of Knee and Ankle Sagittal Plane Joint Restrictions During Squatting. (2005)
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The purpose of this study was to evaluate compensatory biomechanical patterns in the lower extremity created by restricted knee flexion and ankle dorsiflexion when performing squats. Forty two healthy subjects (21 men, 21 women; 22.5 (4.5) years, 73.8 (17.8) kg, 167.5 (12.5) cm) participated in the study. Data were collected using a force plate and a 3-d electromagnetic tracking device for bilateral lower extremity analyses.

Three parallel squats were performed in non braced, right knee restricted and right ankle restricted conditions. Dependent measures were hip, knee and ankle total joint displacement and work done on the hip, knee and ankle during the eccentric portion of the squat. Three repeated measures ANOVAs compared lower extremity kinematics between conditions, while one repeated measure ANOVAs evaluated lower extremity kinetics. Mean hip, knee and ankle ROM was reported, as was sagittal plane work done on the hip, knee and ankle for each condition and limb.

The primary findings of this study indicate hip and ankle flexion displacement significantly decreased in the contralateral (non-braced) limb during the ankle joint restricted condition. Ipsilateral (braced) limb hip, knee and ankle flexion significantly decreased during the knee restricted condition, while ipsilateral knee and ankle flexion decreased during the ankle restricted condition. Lower extremity sagittal plane energetic changes occurred in the ipsilateral knee and ankle when the knee joint was restricted and at the ipsilateral ankle in the ankle restricted condition. Additionally, relative and

absolute shifts in work done on the hip, knee and ankle when compared to the non braced squat were noted.

This study may best serve as a general sagittal plane model for clinicians and coaches to reference when using the parallel squat in patients and athletes with knee and ankle dysfunction. This has practical significance to clinicians as these substitutions in work could result in overuse (secondary) injury to the compensatory site or insufficient loading to the dysfunctional site, rendering it weak and susceptible to additional primary injury or limiting the athletes maximal performance.

KINEMATIC AND KINETIC EFFECTS OF KNEE AND ANKLE SAGITTAL PLANE JOINT RESTRICTIONS DURING SQUATTING

by

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APPROVAL PAGE

This dissertation has been approved by the following committee of the

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CHAPTER I

INTRODUCTION

The squat exercise is commonly used by strength coaches and clinicians because of its biomechanical similarities to sporting activities of running and jumping (Dunn et al., 1984 & Escamilla et al., 1998). This exercise is integral to lower extremity strength enhancement and rehabilitation of injuries to the ankle, knee and hip Shelbourne, 1990 & Fu, 1992; Bynum, 1995). Lower extremity injuries may disrupt normal squatting biomechanics by creating compensatory movements placing otherwise non-injured body segments at increased risk of injury (Salem et al, 2003 & Howard et al., in revision). Clinicians as well as coaches should be concerned that such compensations could lead to reinjury or injury to another body area secondary to excessive or abnormal loading during exercise or sport related activities.

The multi-joint nature of the squat exercise makes it an ideal range of motion (ROM) and integrated strength assessment tool of the ankle, knee, hip and trunk. The squatting motion begins from an erect stance position with the hips and knees fully extended. The descent phase of the parallel squat consists of the ankle, knee, and hip segments moving in bilateral, coordinated sequences maintaining the center of mass (COM) within the base of support (BOS). Hip and knee extensor moments act as coupled movements when squatting because of the effects of the line of gravity with respect to the hip and knee joint centers. A more flexed hip position moves the line of gravity anteriorly, decreasing the knee extensor moment and increasing the hip extensor moment whereas a more vertical trunk (decreased hip flexion) will shift the muscular effort from the hip toward the knee extensors. A hip or knee strategy can selectively influence work across the lower extremity joints. Thus detecting the preferred movement pattern is important to ensure that the exercise is targeting the intended site (Salem et al., 2003).

Although there remains no universal acceptance as to what constitutes the ideal squat, in many circumstances it appears categorically specific. Powerlifters often squat with a wide stance and squat to depths that exceed a parallel thigh to floor position (McLaughlin et al., 1977), while bodybuilders are noted for their use of a variety of stances and depths in an attempt to maximize multiple muscle activation patterns. Moreover, many rehabilitation clinicians advocate a shoulder width stance coupled with shallow knee flexion angles when rehabilitating lower limb injuries (Coqueiro et al., 2005). The recreational exerciser may use any combination of these stance widths and it is the observation of the primary author that most may not achieve a level of knee flexion that optimizes muscular activity across the thighs. Thus, it appears seemingly healthy populations use a variety of squat styles.

Injury to the ankle or knee may compromise normal lower extremity movement when squatting. Up to 25% of athletic injuries involve the foot and ankle complex which in turn may potentially restrict normal ankle motion (McBryde et al., 1997). Decreased ankle dorsiflexion prevents normal anterior tibial motion relative to the talus resulting in altered talocrural movement patterns when performing the squat (Fry et al., 2003). A joint restriction at the knee may also negatively impact squat performance by creating kinematic chain substitutions at the ipsilateral and contralateral ankle, knee and hip (Howard et al., in revision). Having adequate range of motion at the knee and ankle is therefore seemingly essential components to completing the squat correctly.

Lower extremity weakness may also prevent the athlete from moving through a full range of motion when squatting. Muscle atrophy and resulting weakness are expected occurrences with any significant injury or surgery with some studies suggesting strength deficits lasting from up to 49 months post operatively (Lopresti et al., 1988; Arangio et al.,1997; Augustsson et al., 1998 & Salem et al., 2003). Rehabilitation studies that compared multi-joint exercises similar in nature to the squat to single joint exercises like the knee extension, indicate that squat strength can increase without any increase in isolated knee extension strength (Augustsson et al.,1998; Worrell et al., 1996). These findings indicate that compensations for deficits in knee extensor function may exist when using squats as a post operative rehabilitation exercise.

The multiple-joint characteristics of this exercise may permit intralimb substitution patterns that alter effort from the targeted muscle groups (Salem et al., 2003). Moreover, interlimb symmetry may be compromised creating excessive and unwanted load to the contralateral limb and insufficient stimulus to the ipsilateral limb (Howard et al., in revision). These substitution patterns may limit the clinical effectiveness of the squat when used in rehabilitation or strength and conditioning settings. If these compensations persist, a secondary injury is plausible, further disrupting function and athletic performance.

There are few reports on the compensations of biomechanical effects of injury or range of motion restrictions during squatting (Fry et al., 2003; Howard et al., in revision; Neitzel et al., 2002 & Salem et al., 2003;). Subjects post operative Anterior Cruciate Ligament (ACL) reconstruction have been shown to squat with a form that decreases lower extremity moments across the ipsilateral knee when compared to the contralateral knee (Neitzel et al., 2002 & Salem et al., 2003). Bilateral ankle dorsiflexion restrictions have resulted in increased hip moments and decreased knee moments when compared to non restricted squats (Fry et al. 2003). Howard et al. (in revision) unilaterally restricted 15º of knee flexion and reported ipsilateral decreases in hip, knee and ankle sagittal plane range of motion, with center of pressure (CoP) shifting toward the contralateral limb when compared to normal squatting. These studies support the notion that an ankle or knee joint restriction produces an accommodation that may increase neighboring joint demands, resulting in contralateral limb loading or insufficient loading to the restricted joint segment. There are limited studies to date evaluating work demands of the lower extremity joints in a unilateral joint dysfunction during squatting (Neitzel et al., 2002; Salem et al., 2003). This information would help further clinical understanding of how joint restrictions impact loading of the involved and associated lower extremity joints during squatting.

During recovery from ankle and knee injury the squat exercise is used by many clinicians as part of a comprehensive rehabilitation program, which further emphasizes the need to identify compensatory mechanics that may occur as a result of injury (Howard et al., in revision). Because the dynamic squat involves bilateral joint contributions at the ankle, knee and hip, further study of common injury complications such as decreased motion would be helpful to clinicians and coaches in understanding how joint restrictions at the knee and ankle adversely affect loading mechanics and the lower extremity joints during the squat exercise.

Statement of the Problem

Squatting incorrectly may lead to pain, joint impairment, disability, re-injury or a secondary lower extremity injury (Mazur et al., 1993 & Bullock-Saxon et al., 1994). Joint range of motion restrictions are detrimental to performing the squat correctly (Salem et al., 2003; Howard et al., in revision). Ankle and knee joint restrictions may produce distinct compensatory biomechanics that may restrict motion in the ipsilateral limb while excessively loading the contralateral limb (Howard et al., in revision). Knowledge of these compensations will allow coaches and clinicians to specifically modify squat instruction and monitoring strategies when instructing a recovering or "recovered" athlete. The purpose of this project was to evaluate the compensatory biomechanical patterns in the lower extremity created by restricted knee flexion and ankle dorsiflexion when performing squats. This research represents a novel study to attempt to analyze kinematics and kinetics of a joint restricted squat. The hypothesis of this study is that limitations in joint range of motion are a contributing component of altered biomechanics potentially resulting in injury and decreased performance.

Objectives

Objective 1 - Compare squatting kinematics during normal, knee joint flexion restriction, and ankle joint dorsiflexion restriction conditions.

> *Hypothesis 1*: Isolated knee joint and ankle joint dorsiflexion restrictions will produce decreased sagittal plane ROM in the restricted joint when compared to normal squatting conditions.
>
> *Hypothesis 2:* Isolated knee and ankle joint dorsiflexion restrictions wilproduce ipsilateral and contralateral limb substitutions at the ankle, knee and hip during squatting.

Objective 2 - Compare sagittal plane lower extremity energetics at the ankle, knee and hip during squatting in normal, knee joint flexion restriction, and ankle joint dorsiflexion restriction conditions.

> *Hypothesis 3:* Contralateral limb lower extremity energetic demands will be greater during knee and ankle joint restricted conditions when compared to the ipsilateral limb.

Limitations/Assumptions

Restrictions outside of the control of the researcher included the following:

1. Generalizations of the findings may best apply to shoulder width stances with the foot facing directly forward.
2. These subjects represent recreationally active persons who are familiar with performing the squat as part of an exercise regimen and should not be portrayed as representative of ideal form but what may be expected with this population.
3. A mechanical block was used to create the joint restriction so that the restriction would be uniform across participants. An injury resulting in a joint restriction may additionally cause swelling, pain, weakness and proprioceptive deficits. It is not clearly known how or if these additional deficits would cause voluntary or involuntary changes when squatting.

Delimitations

Restrictions imposed in this study by the researcher included the following:

1. Subjects unable to complete the squat trials with satisfactory form were not included in the analysis of this study. Adequate squat form was determined by the primary investigator. See methods (Table 5) for details.
2. Erroneous squat trials were discarded and replaced with acceptable trials.
3. A knee brace was chosen to create a knee joint flexion restriction due to the control that was needed for this study. While it is realized that the brace artificially induces a restriction, the compensation with this task appears

"visually" to mimic what is seen in the rehabilitation clinic when patients present with a variety of knee dysfunctions when attempting to squat. It is also realized that pain, weakness, lack of lumbo-pelvic-hip rhythm and movement variability can also influence the "joint" performance of this exercise. Thus, the results of this study may best be viewed as a theoretical model of compensation that may be present during a percentage of joint dysfunctions at the knee.

4. An ankle restriction was created by preventing the knees from passing in front of the toes via a wooden board. While it is realized that the board artificially induces a restriction, the compensation with this task appears "visually" to mimic what is seen in the rehabilitation clinic when patients present with a variety of ankle dysfunctions when attempting to squat. This method provided the control needed for this study and like the knee may best be interpreted as a theoretical model that may later be validated by prospective analyses of squats in subjects with knee or ankle dysfunction.
5. The squat is a task specific activity that cannot be generalized to other movement patterns.

Operational Definitions

Dorsiflexion (Ankle) - Movement of the foot toward the leg. Average range of motion is 20° (Norkin & Levangie, 1992)

Center of Mass - A balance point of a body; the point at which the body's mass is evenly distributed (Hamill & Knutzen, 2003)

Joint Powers - The product of the moment and angular velocity at a joint (Winter, 1990) Unit = (Newton * Meter) / Second or Watts

Work - The product of the force applied to a body and the distance through which the force is applied or the change in energy of a body (Hamill & Knutzen, 2003) Unit = Newton * meter

Energetics - The study of the change of energy of a body. The ability to do or absorb work. (Robertson et al., 2004)

Joint restriction - The inability to move a joint segment through its full ROM

Ipsilateral - On or referring to the same side. For purposes of this study ipsilateral will refer to the braced limb.

Contralateral – On or referring to the opposite side. For purposes of this study contralateral will refer to the non braced limb.

Squat – A lower extremity multijoint exercise involving the coordinated descent of the limbs to a point where the thigh is in a parallel position to the floor followed by ascent to the upright standing and starting position.

Note:

This study's initial intent was to report multiplanar kinematic analyses of the squat at the hip and knee. Unfortunately, coronal and transverse plane kinematic data appeared to differ from previous squat reliability data (Howard et al., in revision). The reader can find the hip and knee coronal and transverse plane descriptive statistics in Appendix A (SPSS output). It is believed this is related to the left thigh sensor malfunction that resulted in concern of the validity of the coronal and transverse plane kinematic data at the left hip and knee. Thus the decision was made to only compare left and right side sagittal plane kinematics across squat conditions. Since the objective was to compare the changes within the limb across conditions left and right side sagittal plane kinematics were evaluated. The left hip and knee sagittal plane values are similar to previous reliability work examining the effects of knee restrictions during the squat (Howard et al., in revision).

CHAPTER II

LITERATURE REVIEW

The goal of this study is to examine the effects of lower extremity joint restrictions during the squat exercise. This will be done by quantifying sagittal plane kinematics and kinetics at the ankle, knee and hip during sagittal plane restrictions of the knee and ankle. This review of the literature will focus on the following: 1) the importance of the squat exercise in rehabilitation and strength and conditioning, 2) an overview of squatting biomechanics during normal, joint compromised and various stance width conditions, and 3) the benefits of using energetics as a measure of compensatory joint motion.

The Role of the Squat

The squat is a multiple joint exercise that is integral to lower extremity strengthening for performance, injury prevention, and rehabilitation of lower extremity injuries (Bynum et al., 1995; Escamilla et al., 1998; Fleck et al., 1987; Fu et al., 1992 & Shelbourne et al, 1990). Strength coaches often consider the squat essential for maximum development of athletic potential by improving the athlete’s ability to forcefully extend the hips and knees (Fleck et al., 1987). Rehabilitation professionals use this exercise to strengthen the quadriceps and hamstrings and to test and restore function at the lower extremities (Lopresti et al., 1988; Stein et al, 1996; Toutoungi, 2000 & Wilk et al., 1996). Squatting has also been credited for improving performance in biomechanically similar movements such as jumping and Olympic weightlifting (O'Shea, 1985; Palmitier et al., 1991, & Stone et al., 1980). Squats are perhaps the best exercise for preparing the lower extremities for ground based sports. Moreover, they are thought to produce a complete training stimulus due to the balance, coordination, and activation of the lower extremity musculature involved in completing the exercise (McLaughlin, 1977).

Historically, the squat has not been an exercise without controversy. The initial primary concerns were the potential for medial-lateral and anterior-posterior knee instabilities (Klein, 1961 & Klein 1962). Medial and lateral knee stability of 128 healthy competitive weightlifters were prospectively assessed in an effort to quantify the effects of full (posterior thigh contact with calf) squatting. The author reported that full squats led to an immediate increase in medial-lateral as well as anterior-posterior knee laxity. These findings were later disputed over concerns of inadequately described instrument measurement reliability and study methodology (Todd, 1984). Subsequent studies have since reported no difference in knee laxity over short term periods (8 week) when squatting (Chandler et al., 1989 & Myers,1971). Meyers, using a modified version of Klein's instrument reported no differences in collateral ligament stability after squatting (Meyers, 1971). Chandler et al. (1989) examined 100 subjects randomly divided into groups performing the half squat, full squat and one group serving as a control. A KT 1000 knee ligament arthrometer assessed ligament laxity over eight weeks at pre, mid, and post-training intervals with no reported difference in anterior-posterior knee stability between groups. They concluded that the squat is a safe exercise when performed correctly and in fact a deterrent to knee injuries because of the compression increasing joint congruency and stability at the knee. Anterior-posterior knee stability was unchanged in thirty two football players who performed the parallel squat at loads of 130%-200% body weight for 21 weeks using a periodized weightlifting plan (Panariello et al., 1994). These studies support the notion that squats do not increase ligamentous laxity at the knee with no subsequent predisposition to injuries associated with excessive laxity. In fact there is evidence that squatting with multiple sets of 8-12 repetition loads strengthens connective tissues, including muscle, bone, ligament and tendon which in turn may help protect a joint from injurious loads (Chandler et al., 1991 & Stone, 1988). Although earlier in the century there were concerns about the increasing joint laxity during the squat, current literature suggest that the parallel squat is safe and has no negative consequences on medial-lateral or anterior-posterior stability in normal knees (Chandler et al., 1991 & Escamilla et al., 2001).

Tibiofemoral and patellofemoral compressive forces are often mentioned as concerns during squatting. Maximum tibiofemoral compressive forces have been reported to occur between 53-93° of knee flexion however, it remains uncertain how much compression is desirable for a training and rehabilitation stimulus and when excessive compression produces adverse effects at the knee (Escamilla et al., 1998). Patellofemoral malalignment or compressive forces can potentially cause excessive stress on the posterior articular cartilage of the patella resulting in chondomalacia or patellafemoral pain syndrome (PFPS), which is a term used to describe unspecified anterior knee pain (Thomee et al., 1999). In a survey of weightlifters, powerlifters, and non-lifters, knee pain was more prevalent in the lifting groups but less clinical or symptomatic arthritis was reported (Herrick et al., 1983). Furthermore, degenerative changes of the knee have been reported in 15-20% of “senior” lifters who have a history of squatting which is no higher than the age matched general population (Fitzgerald and McLatchie, 1980). While the optimal amount of tibiofemoral compression is undetermined at this time, tibiofemoral compression during squatting may enhance knee stability by reducing anterior knee translation (Chandler et al., 1991; Escamilla et al., 1998; Lutz et al., 1993; Palmitier et al., 1991; Wilk et al., 1996 & Yack et al., 1993).

The National Strength and Conditioning Association (NSCA) developed a set of general guidelines for proper squat form (NSCA Position Paper by Chandler and Stone, 1991). The need for a representative paper examining the role of the squat in athletic conditioning was established by members of the NSCA Research Committee. Jeff Chandler Ed.D, CSCS and Michael Stone, Ph.D, CSCS were the primary authors responsible for the NSCA Position Paper and are considered experts in the field of strength and conditioning. They suggested the following guidelines as reasonable considerations for proper squat form. 1) The lifter should use approximately a shoulder width foot stance when squatting. 2) The lifter should descend in a controlled manner while ascent may be performed at varied speeds with no compromise in technique. 3) Proper breathing supports the core and consists of inhaling from the start of the descent phase through the sticking point of the ascent phase. 4) Proper technique consists of refraining from twisting or bouncing motions at the bottom position of the squat and maintenance of an upright torso with a normal lordotic posture throughout the exercise. 5) Other technical considerations include feet remaining flat on the floor, and minimal forward lean of the knee anterior to the toes (NSCA Position Paper by Chandler and Stone, 1991). These guidelines stemmed primarily from the authors lifting experiences and collective observations.

General consensus from this NSCA committee is that overuse injuries may occur from the squat exercise if proper form and sensible progression according to established exercise program design are not followed. Furthermore it goes on to state that injuries attributed to the squat are likely the result of improper technique, pre-existing structural abnormalities, fatigue, or overtraining (NSCA Position Paper by Chandler and Stone, 1991). When proper form is used the squat is believed to be safe and effective for all healthy populations.

Squat Biomechanics

Squat studies focusing on lower extremity biomechanics have typically reported knee sagittal plane kinematics, joint moments, and variations in stance width. Studies in the field of rehabilitation have evaluated post operative lower extremity bilateral symmetry and strength deficits. Additionally it may be beneficial to consider multiplanar lower extremity kinematic accommodations as the result of joint injury or musculoskeletal weakness. Finally, joint energetics analyses would appear to be beneficial in quantifying the lower extremity's joint responses to squatting and subsequent load sharing. The following sections will review relevant literature and provide support for the examination of ankle and knee joint restrictions and the biomechanical impact they may have on the lower extremities during squatting.

Most studies reporting squat kinematics have done so using two-dimensional analyses reporting only sagittal plane joint ranges of motion (Escamilla et al., 2000; Lander et al., 1986 & McLaughlin et al. 1977). However, there is concern regarding the accuracy of tracking only two-dimensional motion at the lower extremities during shoulder width and wide stance squatting (Escamilla et al., 2000). Wider stance widths cause joint motion to deviate from the sagittal plane compared to narrow stance widths (less than shoulder width) potentially underestimating hip and knee sagittal plane range of motion using two- dimensional analyses. The authors reported shoulder width stance sagittal plane ROM values of 109 ± 8° for hip flexion, 102 ± 7° for knee flexion and 26 ± 4° for ankle dorsiflexion when using three-dimensional analysis for the squat with two dimensional analyses underreporting hip and knee motion by 3-13° (Escamilla et al., 2000).

Experienced powerlifters and weightlifters performed three, one repetition 90-100° knee flexion squat trials with a 12 repetition maximum load (mean 146.5 ± 39.0 kg) using “preferred” mean stance widths of 40 ± 8 cm (inside heel to inside heel shoulder width) and forefoot abduction of 22 ± 11° (Escamilla et al., 1998). The purpose of this study was to quantify knee forces and muscle activity across the squat, leg press and leg extension. Kinematic, kinetic and electromyographic data were calculated for only the left lower extremity with bilateral symmetry assumed. The squat generated approximately twice as much hamstring activity as the leg press or leg extension exercises. This preferred stance width and foot position produced greatest quadriceps muscle activity near full flexion during the squat. Additionally, patellofemoral and tibiofemoral compressive forces were greatest nearing full flexion during the squat. Together, this could have implications in athletes recovering from knee injuries when squatting at 90-100° knee flexion ranges. Given the demands on the knee in the peak knee flexion range during squatting, if a joint dysfunction or weakness exists, it is conceivable that the athlete or patient may compensate in this range.

Since the squat primarily occurs in the sagittal plane the coronal and transverse planes of movement often go unreported. A biomechanical analysis and corresponding theoretical model for the squat consists of descending phase knee flexion, internal rotation of the tibia, subtalar joint (STJ) pronation and ankle dorsiflexion (Tibero (1987) and O'Shea (1985)). During closed chain activities, like the squat, STJ and knee motions are interdependent and the internal tibial rotation that occurs appears as an obligatory action necessary for normal kinematics at the knee and ankle (Greenfield 1993 & Tibero 1987). However, they did not report the stance width or foot out position for the squat which could have numerous implications on their theoretical model (Escamilla et al., 2000).

The only located multiplanar report of kinematics at the hip, knee and ankle is that of Howard et al (in revision) who evaluated normal and joint restricted squat joint ranges of motion. Transverse and coronal planes should be included in studies examining the impact of normal and joint restriction squatting. Analyses of these planes may lead to more descriptive conclusions of lower extremity kinematic changes or accommodations resulting from injury or weakness when compared to only sagittal plane report.

Lower extremity three-dimensional kinematics at the hip, knee and ankle during a shoulder width parallel squat has been reported (Howard et al., in revision). Findings beyond the expected sagittal plane flexion included hip external rotation and knee internal rotation, hip abduction, and knee adduction during the descending phase of the squat (Table 1). Subjects performed the parallel squat on two separate days to allow for between day reliability analyses. This data suggests that frontal and transverse plane movements are subtle components of proper squatting.

Normal, parallel squat between day reliability measures were generally high with the exception of left hip rotation and left knee adduction. Further analysis of the ICC components revealed low between subjects variance in hip rotation resulting in suppression of the ICC. Coronal plane knee motion was less reliable suggesting clinicians scrutinize coronal plane knee motion values when comparing squat trials.

	$ICC_{2,k}$(SEM°)	Mean (SD)°
Total joint displacement (descent)		
Left hip flex	0.70 (6.3)	108.2 (11.0)
Right hip flex	0.70 (6.5)	114.4 (11.3)
Left hip ext rot	0.41 (5.7)	16.8 (7.4)
Right hip ext rot	0.70 (5.9)	-19.0 (10.7)
Left hip abd	0.80 (4.8)	18.1 (9.4)
Right hip abd	0.76 (2.9)	-7.2 (5.9)
Left knee flex	0.70 (4.8)	101.8 (6.1)
Right knee flex	0.80 (3.8)	109.2 (7.8)
Left knee int rot	0.81 (4.8)	-22.2 (10.7)
Right knee int rot	0.76 (4.9)	17.8 (10.2)
Left knee add	0.40 (8.7)	-11.8 (11.6)
Right knee add	0.89 (4.2)	2.3 (13.7)
Left ankle flex	0.62 (2.9)	21.4 (4.0)
Right ankle flex	0.82 (2.0)	21.9 (4.1)

Table 1 - Reliability and means of lower extremity total joint displacement in a normal, parallel squat.

Beyond the scope of purely kinematic analyses of the squat, researchers have investigated joint moments during the task. It is difficult to compare lower extremity moments across studies due to differences in data acquisition methods. While some studies have used a single camera with no force platform (Fry et al., 2004; Mcclaughlin et al., 1978; Nisell & Ekholm, 1986; & Russell& Phillips, 1986), others have used a single camera and a single force platform (Lander et al., 1986; Russell & Phillips 1989, Wretenberg et al. 1996), or multiple cameras and one foot on a force platform (Stuart et al., 1996). Some studies quantified joint moments relative to system weight (barbell and body masses) (Lander et al., 1986; Mclaughlin et al., 1976; Russell & Phillips, 1986;

Stuart et al., 1996; and Wretenberg et al., 1996) while others relative to barbell weight only (Nisell & Ekholm, 1986). Taken together this leads to great difficultly in attaining a consensus as to how the joints are individually loaded.

External loads lifted have ranged from 20-270 kg causing peak hip extensor moments to range between 50 and 300 Nm at the ankle (Escamilla et al., 2000; Lander et al., 1986; McLaughlin et al., 1976, Nisell & Ekholm, 1986) between 100 and 500 Nm at the knee (Escamilla et al., 1998; Lander et al., 1986; Mcclaughlin et al., 1978; Nisell & Ekholm, 1986, Stuart et al., 1996 & Wretenberg et al., 1996) and between 150 and 600 Nm at the hip (Lander et al., 1986; Mcclaughlin et al., 1978; & Wretenberg et al., 1996). These wide ranges in loading and resultant joint moments make it difficult to draw conclusions as to what constitutes "normal" hip, knee and ankle moments during squatting.

Stance Width and Foot Angle Effects

Despite the squats popularity, there does not appear to be a universal stance width and foot position recommendation, although wider stances are generally associated with greater toe out (Escamilla et al., 2001). This is significant in that lower extremity biomechanics may be influenced by stance width and foot position (Escamilla et al., 2001). Few studies have attempted to quantify squat stance width and foot angles during the squat and determine subsequent effects on performance (Escamilla et al., 1998; Escamilla et al., 2000; Mccaw et al., 1999; Nisell & Ekholm, 1986 & Signorile et al., 1995

Changing stance width and barbell load during the squat were reported to influence muscle activity in the gluteus maximus and adductor longus in the lower extremities (McCaw & Melrose, 1999). Gluteus maximus activation was two times greater in a wide stance vs. narrow stance. An explanation for this finding stems from length tension changes in the gluteals when the hip is abducted and laterally rotated, as occurs when starting the squat from a wide stance. To compensate for this reduced muscle length and force production capability, motor units may need to be activated with a higher frequency to generate adequate muscle forces. Increased adductor longus activity was reported during the wide stance position. The authors reasoned this was due to increased abduction during wide stances requiring greater adductor recruitment when compared to narrow stance. Several studies support high levels of quadriceps activity during narrow and wide stance squatting (Escamilla et al., 1998; Mccaw & Melrose, 1999; Ninos et al., 1997; Signorile et al., 1995; Stuart et al., 1996 & Wretenburg et al., 1996) but no significant changes between conditions have been reported (Escamilla et al., 1997 & Mccaw & Melrose, 1999).

Quadriceps muscle activity during squatting with the feet turned medially and laterally has been examined (Signorile et al. 1995 & Ninos et al. 1997). Although it appears adductor and gluteal muscle activation is varied according to stance width, these studies concluded no differences in quadriceps muscle activity when comparing foot positions ranging from 15° inward to facing directly forward to 30° outward.

Joint kinematics and moments have often been reported during squatting (Escamilla et al., 2000; Fry et al., 2004; Lander et al., 1986 and Lander et al., 1990;

Mclaughlin, 1978; Nisell & Ekholm, 1986; Salem et al., 2004 & Wretneberg, 1996) but only Escamilla et al. (2000) has evaluated the influence of stance width and foot angle on joint kinematics and moments at select joint angles. Subjects squatted with three stances, narrow stance, medium stance and wide stance with most biomechanical differences noted to occur between narrow stance and wide stance conditions. At 45°, 90°, and maximum knee flexion angles, there was approximately 10° more hip flexion in the medium stance and wide stance groups compared to the narrow stance. The thighs were 10° more horizontal, whereas the shanks were about 8° more vertical and the feet were turned out about 6° more in the wide stance condition. Relative to ankle dorsiflexion, the knees moved forward over the feet 21.7 ± 4.4 cm during narrow stance, 18.0 ± 2.6 cm during medium stance, and 16.0 ± 4.6 cm during wide stance, leading to significant narrow stance differences compared to medium stance and wide stance groups. In light of the kinematic differences that exist more notably between narrow stance and wide stance, it seems logical to further establish biomechanics of the squat in a standardized medium stance position with the feet facing anterior (0°turn out). This may best serve as a beginning point from which to evaluate squat kinematics when comparing narrow stance and wide stance conditions.

Higher knee extensor moments existed in medium and wide stance compared to narrow stance at 45°, 90° and maximum knee flexion angles, likewise higher hip extensor muscle moments were reported in medium stance and more markedly in wide stance at 45 ° knee flexion suggesting greater gluteal and hamstring hip extensor activity (Escamilla et al., 2000 & Mccaw & Melrose, 1999). Ankle plantar flexor muscle moments were generated during narrow stance conditions, whereas ankle dorsiflexion muscle moments were produced in the medium stance with greatest disparity between narrow and wide stance conditions. Knee extensor muscle moments were higher in medium and wide stance conditions compared with the narrow stance. However, Mccaw & Melrose (1999) refuted any significant differences in EMG quadriceps activity between narrow and wide stances. Similar to the kinematic results, kinetic differences are most disparate between narrow stance and wide stance conditions. Since the aim of this study is to examine how joint restricted conditions affect squats, it seems logical to initially exam medium stance as a means of control for variables such as stance width and foot out angles is a logical starting point. This stance may best serve as a baseline toward future studies examining joint restrictions during squatting.

Lower Extremity Compensation

Hip and knee extensor moments are coupled when squatting because of the effects of trunk flexion on the line of gravity with respect to the two joint centers. A more flexed trunk position moves the line of gravity anteriorly toward the knee, lessening the knee extensor moment and increasing the hip internal extensor moment. Likewise a more vertical trunk (decreased hip flexion) will shift the muscular effort from the hip toward the knee extensors. Therefore a hip or knee dominant strategy can influence work demands across the respective joints having important implications when using the squat for rehabilitating lower extremity injuries.

Few studies have been located evaluated lower extremity biomechanics during squatting under the influence of joint dysfunction. (Augustsson et al., 1998; Neitzel et al, 2002; Fry et al., 2004; Howard et al, in revision & Salem et al., 2003). The following section addresses limited reports of kinematic and kinetic effects of joint restrictions when squatting.

A high percentage of patients who have torn their ACL will undergo surgery and use the squat in post operative rehabilitation plans. Sagittal plane kinematics and kinetics of the ankle, knee and hip joints were assessed during squatting after unilateral anterior cruciate ligament (ACL) reconstruction (Salem 2003). Ground reaction forces, joint excursion angles and hip, knee and ankle peak moments of eight subjects with ACL reconstruction with a mean post operative time of 30 ± 12 weeks were measured.
The peak knee extensor moment generated in the noninvolved limb was 25.5% greater than the involved. The authors additionally reported a trend toward greater hip extensor moments in the involved limb which was supported by a greater ratio (46.5%) of peak hip to knee extensor moment in the involved limb, whereas the noninvolved limb shared the load equally between the hip and knee. The authors concluded that subjects used a compensatory strategy in the involved extremity to reduce efforts at the knee secondary to quadriceps weakness. A potential complication of this compensatory strategy is inadequate training stimuli to the target muscle(s) resulting in persistent weakness across the knee.

Thigh atrophy and weakness are common occurrences after knee injury with lower extremity, bilateral strength deficits are reported to persist up to 49 months post operatively (Augustsson et al., 1998; Lopresti et al., 1988 & Salem et al.,, 2003). Strength training studies using multijoint exercises (such as the squat) but testing isolated quadriceps function indicate that while squat strength increases, isolated quadriceps strength may not (Augustsson et al., 1998 & Worrell et al., 1993). These findings indicate a form of compensation or adaptation may exist when using squats as a post operative rehab exercise to strengthen the quadriceps. This may be due to weakness or subtle errors in squat form due to compromised ROM.

Range of motion restrictions at the ankle or knee joint may compromise normal functional movement resulting in increased loading at neighboring joints during athletic activities, possibly leading to injury (Santos et al., 2003). Bilateral differences in ankle dorsiflexion ROM and hamstring flexibility were reported to be risk factors for overuse leg injury (Soderman et al., 2001). The effects of ankle bracing on hip and knee joint motion during two types of trunk rotation tasks were studied in an effort to understand global effects of limiting one portion of the kinetic chain (Santos et al., 2004). Subjects performed two different left trunk rotation tasks; an open task requiring them to balance on one leg when catching a ball tossed from one of the testers and a closed task requiring subjects to touch a target with their shoulder while keeping their arms relaxed by the sides of their body. The tasks required approximately 70° of collective rotation to complete the tasks. Subjects used more knee internal rotation in the closed task condition in contrast to the open task condition where subjects compensated with upper extremity movement resulting in decreased trunk and knee rotation when braced. While the authors reported a limitation of their study was lack of kinetic analysis limiting further quantification of their results, it supports the notion of a joint dysfunction forcing a compensatory movement from a non restricted site in order to complete a task. This ultimately may have negative consequences of overuse and secondary injury to the compensatory site.

The effects of knee position on hip and knee torques during parallel barbell squats was examined (Fry et al., 2004). The study reported hip and knee joint kinetics when forward displacement of the knee past the toes was restricted by a wooden barrier versus a non restricted condition. For the unrestricted squat, hip torque was 28.2 ± 65.0 Nm and knee torque 150.1 ± 50.8 Nm. For the restricted squat, hip torque equaled 302.7 ± 71.2 Nm and knee torque equaled 117.3 ± 34.2 Nm. The restricted squat condition produced increased hip flexion (more anterior lean of the trunk) with more of a vertical tibia (less ankle dorsiflexion) compared to the normal condition. The vertical tibia was accompanied by greater forward trunk lean which resulted in increased hip moments. In contrast, the normal condition resulted in greater tibial inclination producing higher knee moments and lower hip moments. The authors concluded restricting forward movement of the knees minimized stress at the knee but transferred forces to the hips and low-back region. This could have implications in the rehabilitation of injuries involving the lower extremities. For example, if an athlete attempts to avoid stressing an injured knee they may adapt by using more of a hip strategy to unload the knee.

Lower extremity compensations following ACL reconstruction were assessed by kinetic analyses while subjects performed a single-leg vertical jump and a lateral step-up (Ernst et al., 2000). Hip, knee and ankle extensor moments of 20 ACL reconstructed extremities were compared with 20 uninjured and matched extremities. The aim of this study was to determine whether deficits in the quadriceps femoris muscle to generate extension moments at the knee during a vertical jump or a lateral step up would be compensated by the hip and ankle. Results indicated ACL-reconstructed extremities produced lower knee extensor moments when compared to the controls group during the lateral step up, vertical jump take off and landing. However, there was no difference in summed extension moments (hip + knee + ankle) among extremities during the lateral step up and vertical jump take off conditions. The summed extension moment during the vertical jump landing was less in the ACL reconstructed extremity. The authors concluded the landing deficits may represent inadequate compensation to attenuate eccentric forces which may expose the musculoskeletal structures to injury.

Bilateral lower extremity joint angles and moments were collected for 10 normal subjects and 7 subjects who had undergone an ACL reconstruction (Kowalk et al., 1997). Subjects performed repeated trials of ascending a staircase and power and work were reported across the hip, knee and ankle. The authors compared the normal subjects and ACL deficient patients post-operatively (mean follow up of 6 months). Anterior-posterior knee laxity decreased (7.9 mm to 5.8 mm) while patients functional knee scores increased (70.4 to 88.5). Post operative changes included statistically significant decreases for peak moment (91.9 vs 22.5 Nm), power (181 vs 84 W), and work performed (28.0 vs-5.6J) at the injured knee. These reductions were accompanied by significant increases in contralateral ankle joint excursion, moment and power. The authors concluded the patients in this study were evaluated at 3.2 to 11.3 months post-operatively and were likely still experiencing the acute effects of the reconstruction. This is in agreement with previous studies evaluating quadriceps strength deficits post operatively ranging from 69% of normal at 6 weeks (Rubinstein et al., 1994), 93% of normal at 1 (Tibone et al., 1988) and 85% at 2 years (Inman et al., 1995).

In the interest of including an additional level of study control subjects performed the parallel thigh to floor squat with a 15° right knee flexion joint restriction without external resistance (Howard et al. 2004). Reliability and ROM at the hip, knee and ankle joint and CoP were reported in normal and joint restricted conditions (Table 2 and 3).

Table 2- Reliability of joint restricted lower extremity kinematics

Contralateral limb (left – no brace), Ipsilateral limb (right - braced)

	$ICC_{2,k}$(SEM)°	mean (SD)°
Total joint displacement (descent)		
Contra hip flexion	0.71 (6.9)	107.0 (12.8)
Ipsilateral hip flexion	0.76 (7.1)	107.2 (14.4)
Contra hip external rotation	-0.03 (7.9)	16.5 (7.5)
Ipsilateral hip external rotation	0.56 (7.0)	-19.7 (10.6)
Contra hip abduction	0.87 (3.9)	15.7 (11.2)
Ipsilateral hip abduction	0.87 (4.3)	-8.2 (9.6)
Contra knee flexion	0.47 (4.5)	102.4 (5.9)
Ipsilateral knee flexion	0.77 (3.2)	94.0 (6.07)
Contra knee internal rotation	0.91 (2.9)	-17.0 (11.8)
Ipsilateral knee internal rotation	0.50 (6.9)	18.1 (9.7)
Contra knee adduction	0.72 (5.9)	- 11.4 (11.8)
Ipsilateral knee adduction	0.47 (8.3)	8.42 (11.4)
Contra ankle dorsiflexion	0.63 (3.7)	19.7 (6.1)
Ipsilateral ankle dorsiflexion	0.70 (3.0)	10.7 (4.9)

Table 3 - Reliability of CoP in the normal and joint restricted squat condition (negative value indicates CoP shift to the contralateral limb)

	$ICC_{2,k}$(SEM-cm)	Day 1 mean (SD-cm)	Day 2 mean (SD-cm)
Total Center of Pressure (CoP) *displacement* (descent)			
Normal Squat Cond			
Medial - lateral	0.32 (0.01)	-1.2 (0.8)	-1.0 (0.6)
Joint Restricted Cond			
Medial - lateral	0.95 (0.01)	-2.8 (-3.8)	-4.8 (2.2)

Joint restricted squat between day reliability was moderate to high with the exception of contralateral hip internal rotation, contralateral knee flexion, ipsilateral knee internal rotation and ipsilateral knee adduction. Detailed examination of the data revealed small amounts of variance between subjects for contralateral knee flexion and ipsilateral knee internal rotation trials, therefore this low ICC is not surprising. Contralateral hip rotation was very inconsistent. Further examination of the ICC components revealed a higher amount of error variance than between subject variance, likely due to the difficulty of accurately capturing true physiologic hip motion (Houck et al., 2004). Ipsilateral knee adduction was low due to high error variance.

The right knee joint flexion restriction created by a hinged knee brace consistently produced restrictions in knee flexion. Therefore it was anticipated that the ICC value would be low due to little expected variability between subjects since they were all blocked at 90°. However, neoprene straps used to secure the brace permitted small knee movement within the brace set at the 90° flexion block. This appeared to permit a gradual joint restriction as opposed to a hard block, which may be more practical as it likely better mimics physiologic joint range of motion limitations.

CoP reliability in the medial–lateral direction in the normal condition was low but not surprising as a healthy subject population was not expected to have high variance in this plane during the normal squat trials. The joint restricted ICC value for medial–lateral CoP was high because subjects unloaded the ipsilateral extremity and increased the load in the direction of the contralateral limb suggesting a redistribution of forces across the lower extremity joints.

Overall, subjects were able to perform the restricted squat with equal consistency as that found in the normal squat, supporting its use as a reliable model for simulating and investigating biomechanical effects resulting from range of motion restrictions. Additionally, interpreting transverse plane hip and coronal plane knee motion should be done so with caution due to the difficulty of consistently tracking these motions in normal or restricted conditions.

The joint restricted condition produced increased loading onto the contralateral limb and reduced ipsilateral hip, knee and ankle sagittal plane kinematics when compared to the normal condition (table 4). The joint restriction produced increased ipsilateral hip internal rotation, increased contralateral knee adduction and decreased contralateral knee internal rotation. Although kinetic analyses were not performed it is speculated that stance position, coupled with hip abduction during the descent phase of the squat created a varus moment at the knee causing contralateral knee adduction in the restricted condition.

Table 4 - Lower extremity kinematic differences between normal and restricted conditions Contralateral limb (left - non brace) Ipsilateral limb (right - braced)

	Mean (SD°) normal	Mean (SD°) restricted	P value
Total joint displacement. (descent)			
Contra hip flexion	108.2 (11)	107.0 (12.8)	$F_{(1,17)}$ =1.21 ; P =0.287
Ipsilateral hip flexion*	114.4 (11.3)	107.2 (14.4)	$F_{(1,17)}$ = 15.60; P $\leq$ 0.001
Contra hip external rotation	16.5 (7.4)	16.5 (7.5)	$F_{(1,17)}$ = 0.24; P = 0.631
Ipsilateral hip external rotation*	-16.0 (10.7)	-19.7 (10.6)	$F_{(1,17)}$ = 6.58; P = 0.020
Contra hip abduction	18.1 (9.4)	15.7 (10.6)	$F_{(1,17)}$ = 1.12; P = 0.306
Ipsilateral hip abduction	-7.1 (6.0)	-8.2 (6.2)	$F_{(1,17)}$ = 0.235; P = 0.637
Contra knee flexion	101.8 (6.1)	102.4 (5.9)	$F_{(1,17)}$ = 1.51; P = 0.236
Ipsilateral knee flexion*	109.2 (7.84)	94.0 (6.07)	$F_{(1,17)}$ = 138.36; P $\leq$ 0.001
Contra knee internal rotation*	-22.2 (10.7)	-17.0 (11.8)	$F_{(1,17)}$ = 12.10; P =0.003
Ipsilateral knee internal rotation	17.8 (10.2)	18.1 (9.7)	$F_{(1,17)}$ = 0.001; P = 0.935
Contra knee adduction*	-11.8 (11.6)	-11.4 (11.8)	$F_{(1,17)}$ = 6.46; P =0.021
Ipsilateral knee adduction	2.3 (13.7)	8.4 (15.0)	$F_{(1,17)}$ = 3.81; P = 0.068
Contra ankle dorsiflexion	21.4 (4.0)	19.7 (6.1)	$F_{(1,17)}$ = 3.99; P = 0.062
Ipsilateral ankle dorsiflexion*	21.9 (4.1)	10.7 (4.9)	$F_{(1,17)}$ = 66.45; P $\leq$.000

*Significant differences between conditions

The rationale for studying asynchrony of joint movements is based on the notion of overuse injury. Tiberio (1987) theorized if pronation of the subtalar jont is prolonged beyond midstance during gait, tibial internal rotation will be prolonged. This may result in a mechanical dilemma at the knee, as knee extension begins around midstance and is coupled with tibial external rotation in order to maintain tibiofemoral joint congruity. However, if the tibia is in prolonged internal rotation, the femur must excessively rotate internally to achieve the relative knee external rotation needed for knee extension. This compensatory femoral internal rotation was suggested to alter normal patellofemoral alignment causing excessive contact pressures at the lateral facet of the patella. These compensations may produce changes in lower extremity joint motion and weight distribution that may lead to excessive loading of uninvolved structures. It is conceivable that the squat has even greater effects on tibial and femoral biomechanics considering the greater magnitude of loaded knee flexion and transverse plane range of motion requirements compared to gait. This may have negative consequences leading to secondary injury or result in insufficient stimulus to the targeted site for optimal recovery. Knowledge of substitution biomechanics has implications toward rehabilitation professionals and strength coaches who use squats for patients recovering from knee related diagnoses. Further studies should include appropriate kinetic analyses for a more robust interpretation of results.

Lower Extremity Inverse Dynamics

The squat requires interdependent action of the lower extremities musculoskeletal system in order to overcome external forces and maintain a stable system. A method that has been used in landing and injury prevention research is reporting joint energetics as representative variables of how the hip, knee, and ankle musculature contribute to overcome these forces (Butler et al., 2003; Devita et al., 1992 & Zhang et al., 2000). The following sections will provide: 1) a brief review of the variables necessary for inverse dynamic calculations and 2) lower extremity powers (energetics) as they apply to the squat.

Inverse Dynamic Calculations

Individual joint forces and moments can be calculated through an inverse solution (Winter, 1990). The information required to appropriately calculate joint reaction forces and muscle moments are kinematic or position data, anthropometric measures and force data.

Kinematic (position) data refer to the joint positions of the limb segments of interest within the testing space. These data are commonly acquired by video analysis or electromagnetic tracking systems. Electromagnetic tracking systems have been used to acquire position data by attaching individual sensors to the bony segments to be tested (Blackburn, 2002 & Perie, 2003). Position data of the joint segments are made possible by establishing a global and local coordinate system. The global coordinate system is defined by a fixed orthogonal (X,Y,Z) axis system that establishes the 3-dimensional environment that movement occurs. A local coordinate system for each body segment is then used to establish the segment's location (Z, Y, X) and orientation (rotation around each Z, Y, and X axis) with respect to the global coordinate system (Allard et al., 1995). Lower extremity (squat analysis) sensor placement typically includes bilateral feet, shanks, femurs and the sacrum. Once sensors are secured the process of digitizing joint segments occurs. This includes precise marking of the desired joints to be measured by placing a sensor attached to a stylus to the joints of the foot, ankle, knee and hip. The sensors act as receivers in the electromagnetic field relative to the digitized joints, thus motion can be tracked.

Anthropometric data estimates limb segment mass, length, and joint center locations relative to anatomical landmarks (Demster, 1995 & Winter, 1990). This is based on the premise each body segment has unique masses and lengths that are based on percentages of a person's height and mass.

Ground reaction forces are the most common forces acting on the body. These forces are three-dimensional and resolved into vertical and two shear components (anterior-posterior and medial-lateral directions). These forces must act on a point referred to as the CoP, only then will one have all of the forces necessary for the inverse solution (Winter, 1990).

Once kinematic, anthropometric, and force data have been acquired, joint moments can be calculated. Joint moments represent the internal moment (muscle and ligament) of the given joint to overcome ground reaction and external forces imposed on the joint (Winter, 1990). Joint power (moment * angular velocity) can then be determined from calculations of internal joint moments. Finally, joint energetics (work) can be calculated indicating how the body's musculoskeletal system produces and absorbs energy (Winter, 1990).

Lower Extremity Joint Energetics

Lower extremity performance and injury prevention studies have quantified the energetics of sit to stand concentric performance, protective squat responses during falls, jumping, and landing forces (Butler et al., 2003; Devita et al., 1992; Flanagan et al., 2003; Petrella et al., 2003, Robinovitch et al., 04 & Zhang et al., 2000). At current time there is limited data regarding lower extremity joint energetics during a normal and pathologic squat (Flanagan et al., 2003).

Peak ankle, knee and hip sagittal plane flexion angles, extensor moments, total extensor impulse, joint powers and total work obtained over three trials were reported in an effort to quantify ascending and descending phases of the squat (Flanagan et al., 2003). This is the only study found evaluating energetics when squatting to approximately 100° knee flexion without external loading. Average peak sagittal plane extensor and plantar flexor power ($W \cdot kg^{-1}$) for the hip, knee and ankle ascending and descending phases were reported. Hip power: 0.52 ± 0.21 descending, 0.60 ± 0.25 ascending; knee power: 0.77 ± 0.36 descending, 0.82 ± 0.40 ascending; ankle power: 0.18 ± 0.08 descending, 0.20 ± 0.11 ascending. Total extensor work values ($J \cdot kg^{-1}$) at the hip, knee and ankle were 0.90, 1.25 and 0.25 respectively, indicating that the knee was the primary joint responsible for overcoming squat external forces. There were no statistically significant differences in average peak sagittal plane extensor and plantar flexor power for the hip, knee and ankle between ascent and descent phases. Subjects self selected their stance and speed of movement which makes interpretation of these findings difficult to generalize across populations.

Eccentric control of the lower extremity muscles during the descent phase of the sit to stand (squat) is thought to optimize performance and minimize injury (Schot, 2004). The effectiveness of the squat response in reducing vertical impact velocity was determined through absorption of energy in the eccentrically contracting muscles spanning the ankle, knee and hip (Robinovitch et al., 2004). Hip (76 ± 44 J), knee (53 ±

26 J) and ankle (6 ± 5) work indicated that the proximal joint segments were preferential in reducing impact velocities during a simulated balance test evaluating the effects of squatting when falling backwards.. Collectively these studies support the notion that energetics have an important role in quantifying and better understanding squatting motions. Moreover, segmental (hip, knee and ankle) joint powers in normal and joint restricted squats would be beneficial in indicating which joint segments primarily contribute to mechanical work during standardized parallel thigh squats and how a unilateral joint restriction effects lower extremity joint loading.

In summary, segmental joint energetics enables the researcher to assess which joint preferentially contributes to resolving external forces during the squat. The mechanical power or the rate of energy absorbed and produced on these lower extremity muscles, (Winter, 1990) reflects the magnitude of loading at each joint and thus may indicate the joint's potential for compensation during squatting.

Summary

The goal of this review of literature was to provide a framework supporting the role of lower extremity joint restrictions in producing compensatory motion when squatting. This review is intended to provide a background of the importance of the squat in strength and conditioning and rehabilitation. Normal biomechanics and the effects of varying squat stance width and foot out angles were reported. The effects of lower extremity range of motion and strength deficits during the squat were reviewed in an effort to establish the need for in depth kinematic and kinetic quantification via lower extremity energetics. Finally, reporting lower extremity energetics will provide both individual joint and whole-body measures that reflect how forces are distributed and the individuals' response to overcome forces during a joint restricted squat.

CHAPTER III

METHODS

Design

This study followed a repeated measures design. The independent variables consisted of three squat conditions: 1) non braced, 2) knee joint flexion restricted and 3) ankle joint dorsiflexion restricted. The dependent measures were 1) sagittal plane ankle, knee and hip joint displacements and 2) sagittal plane ankle, knee and hip joint energetics. These variables assessed lower extremity biomechanics between squat conditions and were calculated from kinematic and kinetic data acquired through a three dimensional electromagnetic tracking device interfaced with two force plates.

Subjects

Forty-two subjects 21 male, 21 female (mean (SD): 22.5 (4.5) years, 73.8 (17.8) kg, 167.5 (12.5) cm) volunteered and signed a written consent form approved by the University's Institutional Review Board (See Appendix P) prior to data collection. Power was calculated based on previous hip kinematic effect sizes which determined that a sample size of 42 subjects yielded 0.80 power.

$$d = \frac{\mu_1 - \mu_2}{\sigma_{X1-X2}} = \frac{7.2}{12.41} = .58 \qquad n = 2\left(\frac{\delta}{d}\right)^2 = 2\left(\frac{2.80}{.58}\right)^2 = 41.4 = 42$$

The UNCG Institutional Review Board approved this experiment and subjects in the study gave their informed consent. Subjects were recreationally activity at least three times weekly and demonstrated the ability to perform the squat exercise. Subjects were excluded if they had a history of reconstructive hip, knee or ankle surgery, or received treatment for hip, knee or ankle pain in the last 6 months. Five subjects were unable to squat to a parallel thigh to floor position, using acceptable form as described by the National Strength and Conditioning Association (NSCA) (Chandler and Stone, 1991) were not included in the study.

Instrumentation

Three-dimensional kinematics of the hip, knee and ankle were collected at 100 Hz using an electromagnetic tracking system (Ascension Technologies, Burlington, VT) and Motion Monitor software (Innovation Sports Training, Chicago, IL). This system records the position and orientation of sensors (receivers) with respect to a pulsed DC transmitter. This tracking device allows real time data collection and analyses in six degrees of freedom.

An electromagnetic sensor was secured to each subject at the following anatomical sites: junction of C7/T1, sacrum at the S2 level, left and right lateral thighs at mid-thigh, left and right anteromedial tibias, and left and right proximal shafts of the second metatarsals. Sensors were secured with double sided tape and then covered with pre wrap and cloth tape.

Two Bertec Force Plates, Type 4060-nonconducting (Bertec Corporation, Columbus, OH) acquired three forces (Fx, Fy, Fz) and three moments (Mx, My, Mz) sampled at 600Hz. Before the testing session, the force platforms were calibrated per manufacturer's guidelines. This allowed for comparison of left and right lower extremities (Lopresti et al., 1988).

Squat procedure

Stance width was normalized to each subjects biacromial distance (Escamilla et al., 2001). Squats were performed in this stance, with the feet facing directly forward. The knee flexion restriction was created by a knee brace (TROM, DJ Orthopedics, Vista, CA) (Figure 1). The brace was fitted to the right lower extremity and blocked at 90° of knee flexion, which allowed approximately 95 degrees of knee flexion due to the velcro cushion that allows subtle movement. This amount of flexion restriction (approximately 15°) was determined through previous testing to be sufficient to create compensations that are similar to what may be observed clinically when a patient or athlete may have some form of knee dysfunction (Howard et al., in revision). This created the necessary restriction without compromising the ability to perform the task (Howard et al., in revision).

Figure 1 - Knee Brace to create knee flexion restriction

The ankle restriction was created by a wooden board (40cm wide x 60cm tall) secured to a platform and placed anterior to the right ankle. The platform and board were placed just anterior to the dominant limbs great toe. The reliability of a unilateral ankle restriction was previously assessed from the shoulder width stance position by the goniometric function of the Motion Monitor. Pilot data (n = 8) demonstrated 16.1 ± 3.4° ($ICC_{2,k}$ =0.90, SEM ± 0.61°) of right ankle dorsiflexion, leaving the contralateral (non restricted) side free to move throughout its normal course (Figure 2). A parallel thigh squat using a shoulder width stance requires approximately 21.9 ± 4.1° of ankle dorsiflexion (Howard et al., in revision), thus this condition approximates a 5° dorsiflexion restriction. Anecdotally, this degree of motion impairment appears to closely resemble the clinical presentation of patients squatting who have an ankle joint dysfunction.

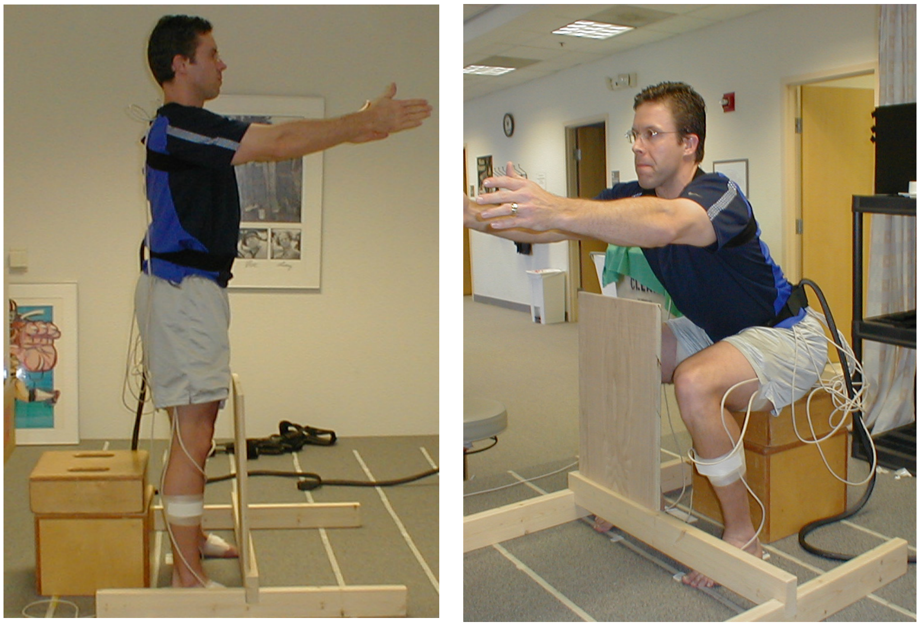

Figure 2 - Ankle dorsiflexion restriction device

Subjects were read a list of instructions and form recommendations prior to the squat (Table 5). Subjects performed body weight squats (no added resistance) and based on previous work, there was little concern of fatigue as a result of performing several practice squats for ensuring comprehension and proper cadence at each condition. When all sensors were secured the subject squatted to a bench height that was adjusted to allow parallel thigh positioning as measured by an inclinometer. The subject achieved slight gluteus maximus contact, but did not relax onto the bench before returning to the upright position. This was determined by the primary investigator during each squat trial through observation. Subjects' arms were outstretched to a parallel to floor position to help maintain balance. A metronome set at 1 Hz ensured a three second descent, one second hold and two second rise thus allowing a uniform and controlled performance between subjects. Condition one was non braced (normal), while conditions two and three were knee and ankle restricted conditions respectively. The squat sequence was counterbalanced between conditions to negate any possible order effect. Three repetitions in each of the three conditions were recorded. Subjects' biacromial width was marked on the force plates to keep subjects' foot position consistent across conditions.

Table 5 - Squat Study Instruction for normal, knee flexion restriction, and ankle dorsiflexion restricted conditions

Squat Study Instruction for **normal** condition

1. Feet will be placed shoulder width apart
2. Feet must remain forward throughout the entire session
3. Your feet must stay in contact with the ground…..(your heels/ toes can not raise up)
4. Sit down and back as if you were going to sit on a chair
5. Let your rear go backwards while simultaneously bending the knees and hinging forward at the hips
6. Once your rear makes slight contact onto the seat surface you may raise back up to the stating position. Do NOT relax onto the seat, only let your rear slightly touch the surface
7. Keep your trunk somewhat upright
8. Your arms will be held out in front of you parallel to the ground to assist you with balance
9. Look straight ahead as you perform the squat task

Squat Study Instruction for the **knee** restricted condition

1. You must follow the above guidelines the best as you can. However, you are allowed to make any subtle adjustments in order to complete the task without loss of balance because the brace will restrict some of your motion.
2. Remember, you must keep your feet stationary and facing directly forward and your rear must only make slight contact with the bench

Squat Study Instruction for the **ankle** restricted condition

1. You must follow the above guidelines the best as you can. However, you are allowed to make any subtle adjustments in order to complete the task without loss of balance because the restriction will limit some of your motion.
2. Remember, you must keep your feet stationary and facing directly forward and your rear must only make slight contact with the bench

Force and electromagnetic tracking equipment were electronically synchronized to sample force data at 600 Hz (Salem et al., 2004) and kinematic data at 100 Hz.

Subjects were positioned with one foot on each force platform allowing data to be analyzed bilaterally.

Data Reduction and Analysis

After all collection was complete, kinematic data were smoothed using a 10 Hz low pass 4th order zero-lag digital Butterworth filter (Winter, 1990). A segmental reference system was used to quantify the kinematics of the lower limb during the squat. Euler's equations were chosen to describe joint motion about the following axes defined in the anatomical segments. The positive mediolateral axis (Z) pointing right, the positive anterior posterior axis (X) pointing anteriorly, and the positive longitudinal axis (Y) pointing superiorly. The order of the rotational sequence used for hip, knee and ankle analysis was (Z,Y', X"). Data for each subject were time normalized creating an ensemble average of the three trials across trials for each condition.

Kinetic data were low passed filtered at 60Hz using a 4th order, zero-lag Digital Butterworth filter. Hip, knee and ankle resultant joint forces and moments from the squat descent phase were calculated from the force platform data and position data using inverse dynamics analyses (Eng & Winter, 1995). All kinematic and kinetic data were then exported into an excel spreadsheet for calculation of the joint energetics. All data considered for analysis was calculated during the descending phase of the squat. The squatting descent phase was operationally defined as starting from an upright standing position (highest total body center of mass) and ending when the total measured body center of mass is at the lowest position relative to the force plate.

Total joint displacements of the hip, knee and ankle were defined as changes in joint angle from initiation of the descent phase to the peak of the descent phase as defined by the most inferior position of the total body center of mass (COM) calculated from the position data of the eight segments measured. Average ipsilateral and contralateral hip, knee and ankle sagittal plane joint displacements were recorded across conditions.

Total work absorption for each of the lower extremity joints were calculated by taking the time integral under the joints respective power curves during the descent phase of the squat (Winter, 1990). The area under the power curve represents the work done on the joints. Joint powers were calculated as the product of the internal joint moment times the angular velocity. Joint powers were normalized to each subject's body mass in kilograms.

To assess the kinematic differences within the ipsilateral (braced) and contralateral (nonbraced) limbs between conditions three repeated measures ANOVA's (condition (3 levels – non braced, knee braced, ankle braced) by limb (2 levels – ipsilateral limb, contralateral limb) were performed on the dependent measures of hip, knee, and ankle range of motion. A three-way ANOVA [condition (3 levels – non braced, knee braced, ankle braced) by limb (2 levels – ipsilateral (braced), contralateral (non braced) by joint (3 levels – hip, knee, ankle)] tested for energetic differences. Follow up two way ANOVA's of condition x limb were performed on hip, knee, and ankle energetics. An alpha level of $P < .05$ was used for all analyses. Tukey's test was used to post hoc test all significant F values.

CHAPTER IV

RESULTS

Kinematics

Hip, knee and ankle sagittal plane descriptive statistics are located in Table 6 The ANOVA performed on hip joint range of motion (ROM), demonstrated a significant interaction between squat condition and limb ($F_{(2,82)}$= 7.082, p<.001, see Appendix C for SPSS outputs) with a significant main effect on limb (P<.001), see table 6 for effect sizes (ES). The Mauchly Test of Sphericity was significant (p<.001), therefore the Huynh-Feldt Epsilon correction was applied in order to protect against Type 1 error. This did not change the condition by limb interaction ($F_{(1.6, 64.7)}$=7.08, p<.003). Tukey's HSD Post-Hoc comparisons of normal to joint restricted conditions identified ipsilateral hip joint displacement decreased [2.4°, ES = 0.16] in the knee restricted condition, whereas, contralateral hip flexion decreased [2.3°, ES = 0.15] in the ankle restricted condition (see Appendix F for calculations). Graphs indicating these changes can be viewed in Figure 3.

Table 6. **Contralateral (non-braced) and ipsilateral (braced) limb sagittal plane total joint displacement means and standard deviations during the descent phase of the parallel thigh to floor squat**: Effect size (ES) for main effect specified as *limb, **condition, °condition and limb, and †significant changes between non braced & knee restricted and non braced & ankle restricted conditions.

	Non braced (ROM)	Knee Restricted (ROM)	Ankle Restricted (ROM)
Contralateral hip	103.7° ± 13.2°	103.1° ± 12.6°	†101.4° ± 13.2°(ES: 0.15)
Ipsilateral hip	**113.2° ± 11.9°**	†**110.8° ± 13.2° (**ES: 0.16)	**112.7° ± 13.4°**
*Main effect on limb	*Limb ES: 0.71	*Limb ES: 0.61	*Limb ES: 0.91
Contralateral knee	98.7° ± 8.9°	98.9° ± 8.9°	97.9° ± 8.1°
Ipsilateral knee	**106.8° ± 8.8°**	†**93.8° ± 6.8°**	†**99.8° ± 9.0°**
**Main effect on cond.		†, **Cond ES: 1.72	†, **Cond ES: 0.77
Contralateral ankle	22.5° ± 5.7°	21.3° ± 5.8°	†20.7° ± 5.2°
Ipsilateral ankle	**22.7° ± 5.7°**	†**17.4° ± 5.6°**(ES: .40)	†**16.2° ± 3.4°**
°Main effect on cond and limb		°Cond ES: 1.05 °Limb ES: 0.69	†, °Cond ES: 0.36 , 1.47 °Limb ES: 1.12

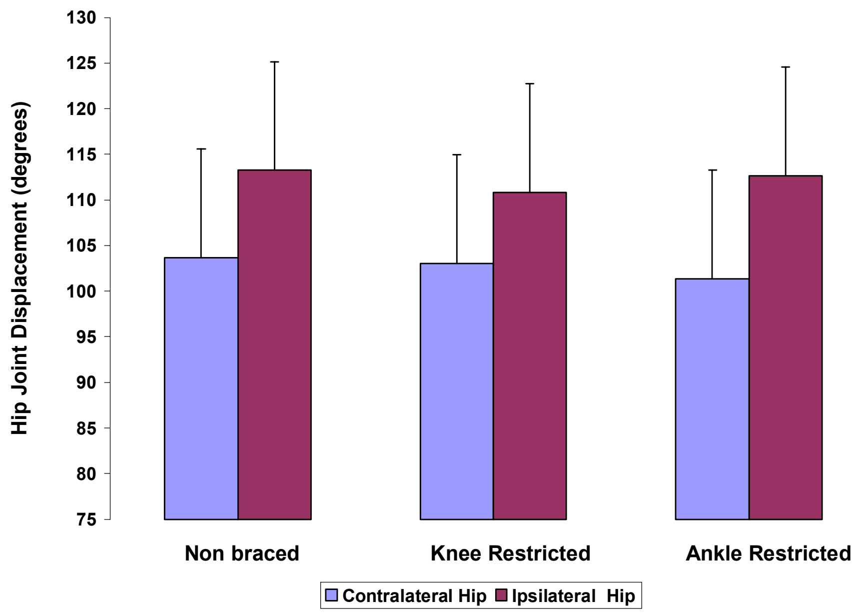

Figure 3. Changes in Total Hip Joint Displacement during Non braced, Knee Restricted and Ankle Restricted Conditions:*†Condition by limb significance, P<.003; †contralateral hip flexion decreased between normal and ankle restricted condition, whereas *ipsilateral hip flexion decreased between non braced and knee restricted condition.

The repeated measures ANOVA performed on knee joint range of motion (ROM), demonstrated a significant interaction between squat condition and limb ($F_{(2,82)}$= 77.73, P<.001, see Appendix D for SPSS outputs) with a significant main effect on condition (P<.001), see table 6 for ES. Means and standard deviations are presented in Table 6. The Huynh-Feldt correction applied to the condition by limb interaction did not change significance ($F_{(1.6, 66.1)}$=77.73, p<.001). Tukey's Post Hoc comparisons identified that when compared to the non brace condition, ipsilateral knee displacement decreased [13.8° (ES=1.72)] in the knee restricted condition and decreased [7.1° (ES = 0.77)] in the ankle restricted condition while there was no change in the contralateral knee (See Appendix G for calculations). Graphs of these changes can be seen in Figure 4.

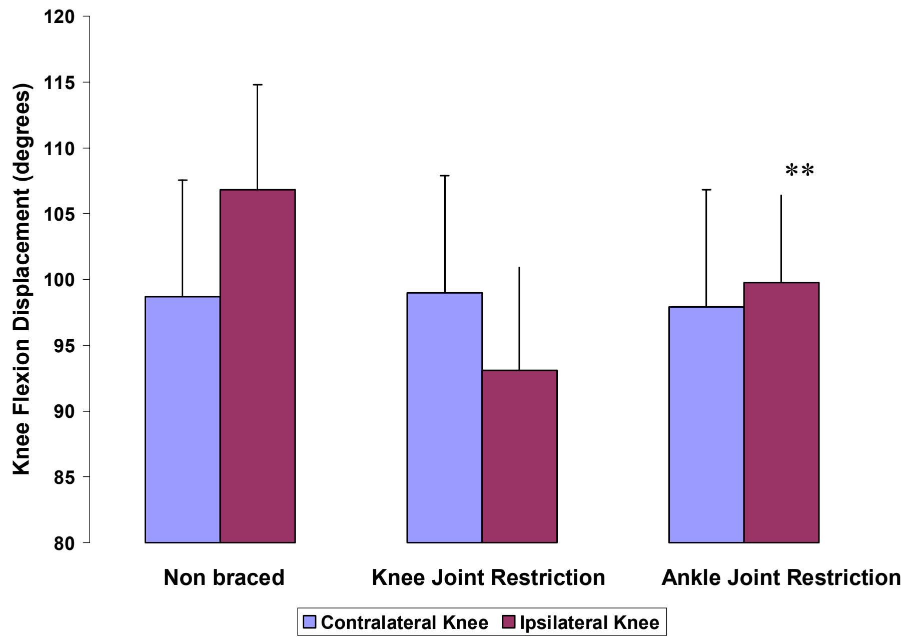

Figure 4. Changes in Total Knee Joint Displacement during Non braced, Knee Restricted and Ankle Restricted Conditions:* **Condition by limb significance, P<.0001; *ipsilateral knee flexion decreased between non braced and knee restricted conditions and ** non braced and ankle restricted conditions.

The repeated measures ANOVA performed on ankle joint range of motion (ROM), demonstrated a significant interaction between squat condition and limb ($F_{(2,82)}$= 35.149, $P<.001$, see Appendix E for SPSS outputs) with significant main effects on condition ($P<.001$) and limb ($P<.001$). The Mauchly's Test was not significant, thus no correction for the degrees of freedom was necessary. Means and standard deviations are presented in Table 6. Tukey's Post –Hoc comparisons identified ipsilateral ankle ROM decreasing [5.3° (ES = 1.05)] in the knee restricted condition and [6.6° (ES = 1.47] in the ipsilateral ankle restricted condition when compared to the no-brace condition (See Appendix H for calculations). Contralateral ankle ROM decreased [1.8° (ES = .36)] when the ankle was restricted. Graphs of these changes can be seen in Figure 5.

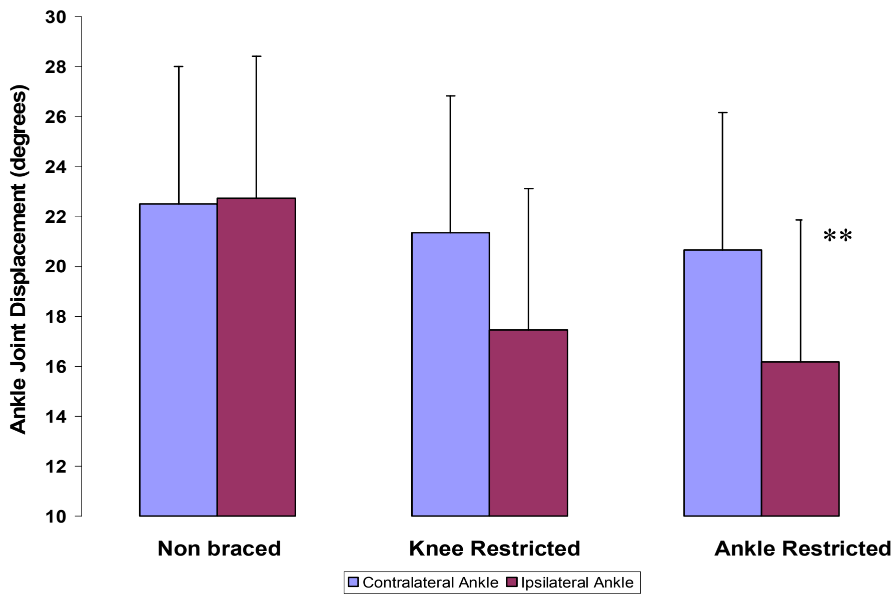

Figure 5. Changes in Total Ankle Joint Displacement during Non braced, Knee Restricted and Ankle Restricted Conditions: * ** †Condition by limb interaction, P<.001; *ipsilateral ankle dorsiflexion significantly decreased between non braced and knee restricted conditions and **non braced and ankle restricted conditions. †Contralateral ankle dorsiflexion significantly decreased only in the ankle restricted condition.

Energetics

A three way interaction of condition by limb by joint indicated significant differences in the work done on the lower extremity joints ($F_{(4,164)}$= 7.203, P<.001, see Appendix I for SPSS outputs). A graph showing this interaction can be viewed in Figure 6 with descriptive statistics found in table 7. The Huynh-Feldt correction was applied to the interaction secondary to the significant Mauchly's Test of Sphericity ($F_{(2.6,107)}$= 7.203, P<.001). As with the kinematics, the correction factor produced no changes from the sphericity assumed values in any condition.

Table 7. Contralateral (non-braced) and ipsilateral (braced) limb sagittal plane energetics means, standard deviations, and relative work contributions during the descent phase of the parallel thigh to floor squat: Effect size (ES) for main effect specified as is the effect size for *limb, **condition, °condition and limb, and †significant changes between non braced & knee restricted and non braced & ankle restricted conditions.

	Non braced Nm/kg	Knee Restricted Nm/kg	Ankle Restricted Nm/kg
Contralateral hip	0.13 ± .08 (22%)	0.14 ± .11 (23%)	0.11 ± .08 (19%)
Ipsilateral hip	**0.21 ± .19 (28%)**	**0.22 ± .18 (32%)**	**0.20 ± .20 (30%)**
*Main effect on limb	*Limb ES: 0.54	*Limb ES: 0.52	*Limb ES: 0.58
Contralateral knee	0.43 ± .12 (74%)	0.45 ± .14 (75%)	0.45 ± .13 (79%)
Ipsilateral knee	**0.49 ± .23 (66%)**	**†0.40 ± .20 (59%)**	**0.44 ± .21 (66%)**
**Main effect on cond		†,**Cond ES: 0.78	
Contralateral ankle	0.02 ± .02 (4%)	0.01 ± .01 (2%)	0.01 ± .01 (2%)
Ipsilateral ankle	**0.04 ± .05 (6%)**	**†0.06 ± .05 (9%)**	**†0.03 ± .04 (4%)**
°Main effect on cond and limb	°Limb ES: 0.79	°Limb ES: 1.36 †,°Cond ES: 0.40	°Limb ES: 0.67 †°Cond ES: 0.22

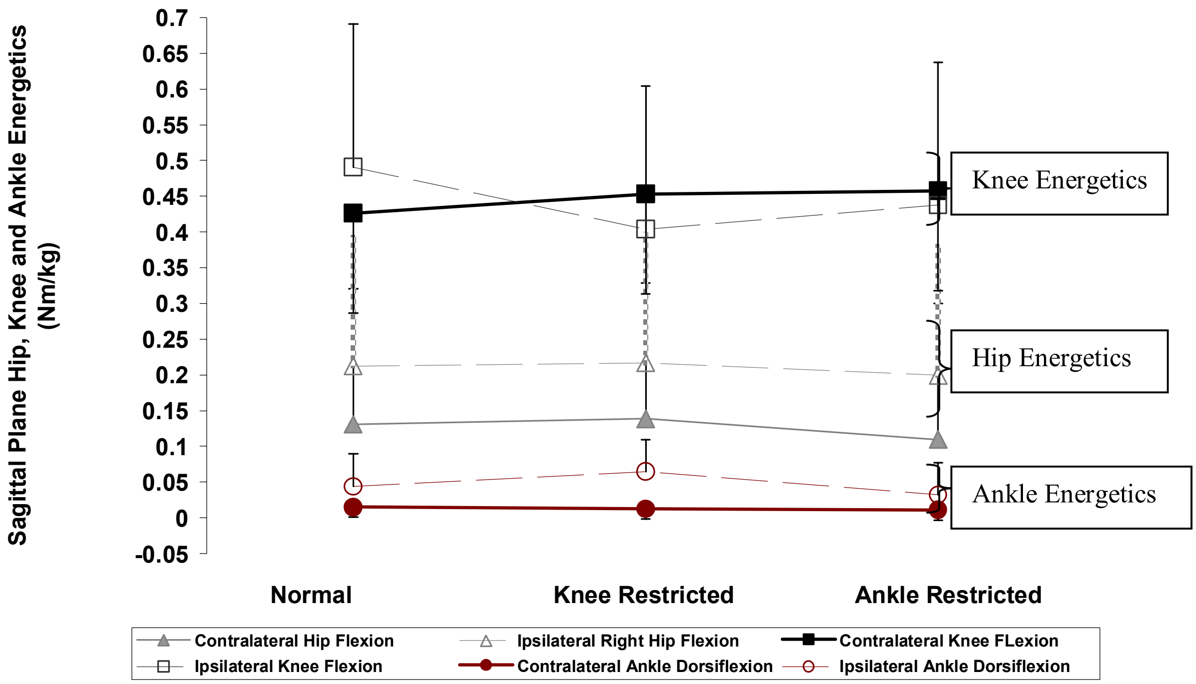

Figure 6. Work done on the ipsilateral (braced) and contralateral (non braced) hip, knee and ankle during the descending phase of the squat across non braced, knee restricted and ankle restricted conditions: *Joint by condition by limb interaction, $P<.001$; *work done on the ipsilateral knee decreased between normal and ipsilateral knee restricted conditions; Significant main effects were noted for condition ($P=.05$), limb ($P<.001$) and joint ($P<.001$). In order of magnitude, work was greatest at the knee > ankle> hip.

Follow up two way ANOVAs of condition x limb were performed on hip, knee, and ankle energetics to better interpret the three way energetic interaction (see Appendices J, K & L for SPSS outputs). The Huynh-Feldt correction was only applied to hip and knee ANOVA's as the sphericity assumption for the ankle was met. The condition by limb interaction was not significant at the hip ($F_{(2,82)}$ =.113, P=.893, Huynh-Feldt correction: $F_{(1.8, 73.9)}$=.113, P=.874, see Figure 7 for graphical display) although a significant main effect was observed at the hip across limbs (P=.005), see table 7 for ES.

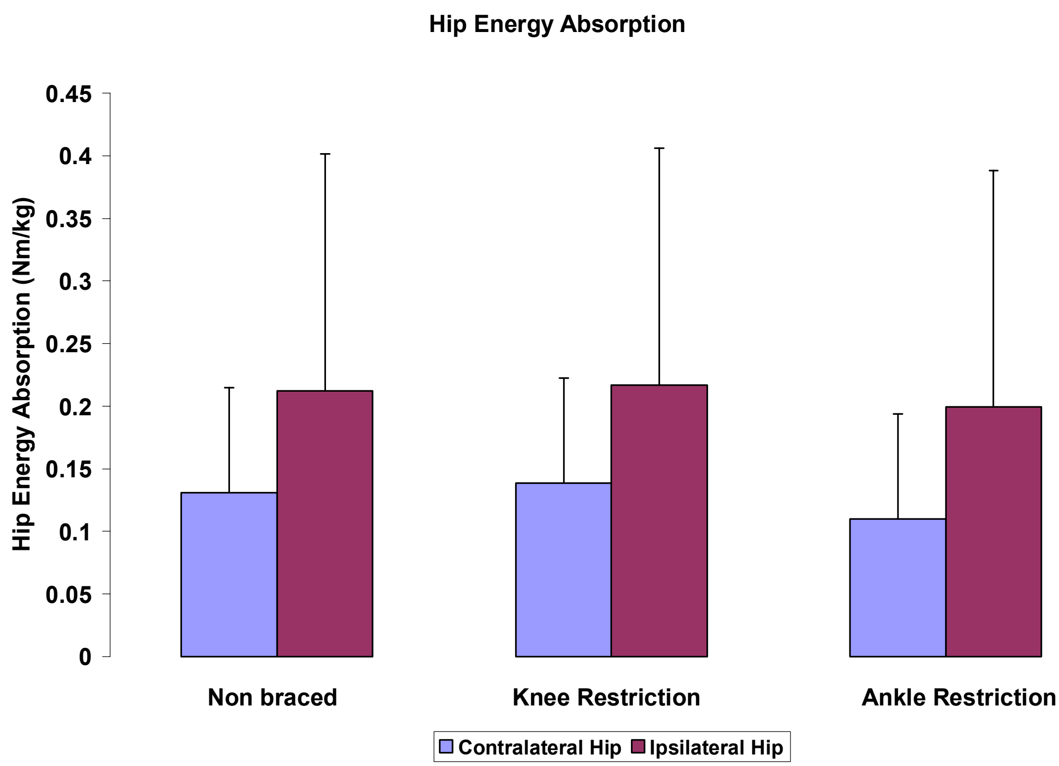

Figure 7. Work done on the ipsilateral (braced) and contralateral (non braced) hip during the descending phase of the squat across non braced, knee restricted and ankle restricted conditions: Significant main effects across limbs (P=.005).

The condition by limb interaction was significant at the knee ($F_{(2,82)}$= 17.53, P< .001, Huynh-Feldt correction: $F_{(1.6, 67.3)}$= 17.53, P< .001, see Figure 8 for graphical display) with a significant main effect on condition (P<.05), see table 7 for ES.. Tukey's HSD Post Hoc comparisons showed the knee restriction significantly reduced the work done on the ipsilateral knee [-0.07 Nm/kg, ES = .[illegible]* compared to the non braced condition (see Appendix N for calculations).

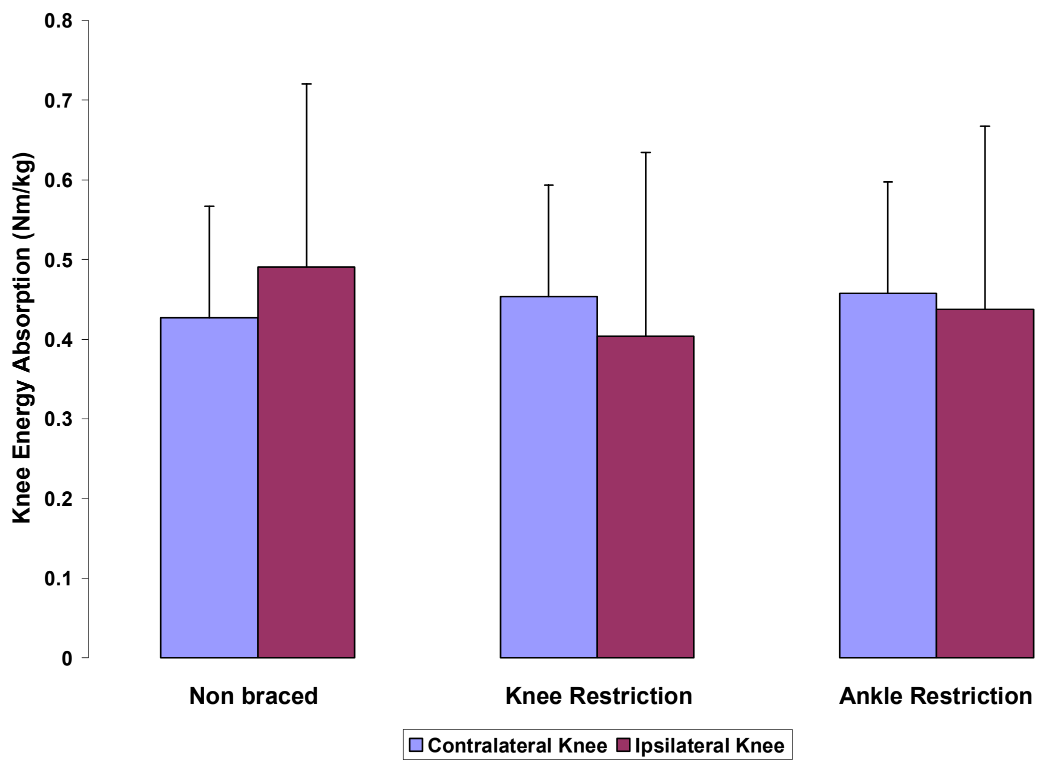

Figure 8. Work done on the ipsilateral (braced) and contralateral (non braced) knee during the descending phase of the squat across non braced, knee restricted and ankle restricted conditions: *condition by limb interaction, P<.05; *work done on the ipsilateral knee decreased between non braced and ipsilateral knee restricted conditions. .

The condition by limb interaction was significant at the ankle ($F_{(2, 182)}$=18.52, P=.001, see Figure 9 for graphical display) with a significant main effect on condition (P<.001) and limb (P<.001), see table 7 for ES. Mauchly's Test was not significant. Post Hoc testing revealed the knee restricted condition resulted in increased work at the ipsilateral ankle (ES = .39), while the ankle restriction decreased work at the ipsilateral ankle (ES = .22) (see Appendix O). There were no significant within limb changes in the contralateral limb.

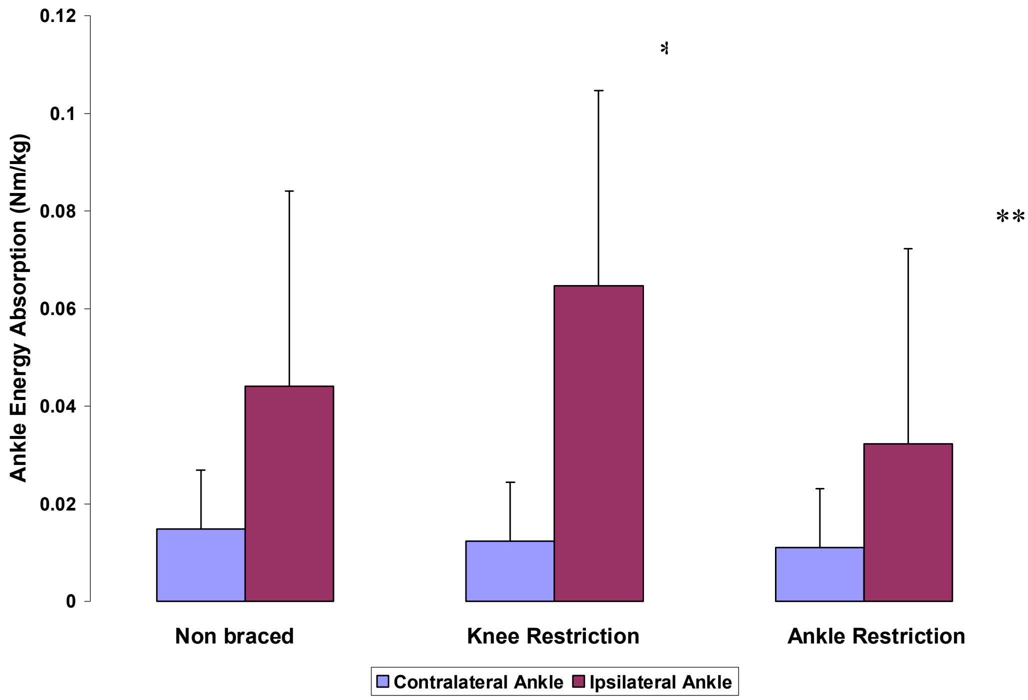

Figure 9. Work done on the ipsilateral (braced) and contralateral (non braced) ankle during the descending phase of the squat across non braced, knee restricted and ankle restricted conditions: * **Condition by limb interaction, P<.001; *work done on the ipsilateral ankle increased between non braced and ipsilateral knee restricted conditions, whereas **work decreased at the ipsilateral ankle.

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CHAPTER V

DISCUSSION

The primary findings of this study indicate that hypothesis 1 was accepted as external applied joint restrictions decreased the restricted joints' ROM. Additionally, hypothesis 2 was partially accepted as hip and ankle flexion displacement significantly decreased in the contralateral (non-braced) limb during the ankle joint restricted condition. Ipsilateral (braced) limb hip, knee and ankle flexion significantly decreased during the knee restricted condition, while ipsilateral knee and ankle flexion decreased during the ankle restricted condition. Finally, hypothesis 3 was not accepted although lower extremity sagittal plane energetic changes did occur in the ipsilateral knee and ankle when the knee joint was restricted and at the ipsilateral ankle in the ankle restricted condition.

Ipsilateral and Contralateral Sagittal Plane Squat Kinematics

A general assumption pertaining to this study was that the joint restrictions created by the knee brace and wooden board resemble the squat pattern anecdotally seen in a variety of knee and ankle injuries that may alter an individual's normal squat style. The primary objective was to compare ipsilateral and contralateral sagittal plane within limb squat kinematics during non braced, knee joint restricted and ankle joint restricted conditions.

Currently there is a paucity of literature evaluating the effects of lower extremity joint dysfunction during squatting (Augustsson et al., 1998; Neitzel et al., 2002; Fry et al., 2004; Howard et al, in revision; & Salem et al., 2003). These authors have generally concluded that knee or ankle joint dysfunction results in knee extensor moment deficits and/or a ROM reduction in the restricted joint. The current study reveals similar results as the knee restricted condition resulted in significant decreases in ipsilateral flexion not only at the knee (13.0°), but also at the hip (2.4°), and ankle (5.3°) when compared to the non restricted condition. This is in agreement with Howard et al. (in revision) who previously demonstrated a knee restriction having similar effects on ipsilateral sagittal plane kinematics further supporting this study's hypothesis that joint restrictions result in decreased ROM at neighboring joints in the ipsilateral limb. Additionally, contralateral limb hip (2.3°) and ankle (1.8°) flexion significantly decreased in the ankle restricted condition which supports this study's hypothesis of contralateral limb effects during a joint restricted squat. There was no significant sagittal plane change in the contralateral limb during the knee restricted condition.

A closer examination of the kinematic data reveals important considerations in the clinical setting when selecting the squat exercise for an athlete or any other population recovering from knee or ankle dysfunction that may alter normal kinematics at that joint segment. The double leg squat is dependent on both extremities for proper execution, thus a joint dysfunction regardless of the source (ie. injury, muscular, ligament, pain or weakness) may conceivably alter normal squat form. Although not a primary focus of this study moderate to large effects were noted at the hip and knee between limbs in the non braced condition suggesting a limb dominance effect during the parallel squat with this population. The discussion will next examine joint specific restriction effects on the entire lower extremity.

Knee Joint Specific Restrictions

The intent of the joint restrictions in this study was to visually replicate what is often seen clinically, regardless of the contributing factors causing compensations. Thus, a brace with set flexion stops was chosen to induce a relatively uniform mechanical restriction across subjects that would result in similar limitations across subjects. The amount of knee flexion produced in the non braced squat condition was 106° ± 8.8°, compared to the knee restricted condition producing 93° ± 7° of knee flexion resulting in large effects (1.71). Thus, the knee restriction successfully decreased ~13° of ipsilateral knee flexion compared to the non braced condition. Furthermore, this magnitude of change appears to be effective in visually replicating shifts seen in a variety of knee conditions seen clinically.

The hypothesis of ipsilateral hip and ankle flexion decreasing when unilateral knee flexion is restricted was supported by the current findings. Decreased ipsilateral hip flexion signifies a proximal joint effect due to a distal knee restriction. Thus, the knee restriction appears to produce a proximal compensatory motion restriction at the hip when compared to the non braced condition suggesting a kinetic chain relationship within the ipsilateral limb.

The kinetic chain has been described as a concept in which the ankle, knees, and hips act as a link system capable of dissipating and transmitting forces into the pelvis and spine during functional activities (Nicholas et al., 1977). In addition to describing how the joints of the lower limb work together to transfer forces between limb segments during motion, biomechanical studies have insinuated that any compromise of a joint segment may lead to dysfunction elsewhere within the extremity (Clement et al., 1984; Teitz et al., 1987; Devita et al., 1992 and Nadler et al., 1998). Thus, kinetic chain concerns may arise secondary to an existing lower extremity injury and/or inadequate rehabilitation of those injuries, as both are established risk factors for future lower extremity injury (Ekstrand & Gillquest, 1983; Agre & Baxter, 1987). In addition, changes in strength and ROM may result from lower extremity injury, and, in accordance with the kinetic chain concept, these changes may occur both proximally and distally to the original site of injury (Agre & Baxter, 1987)

Also in the knee restricted condition, contralateral limb ankle ROM significantly decreased compared to the normal condition albeit a small effect (.34). Decreased contralateral ankle dorsiflexion signifies a distal joint effect of the proximal knee restriction. Thus, ipsilateral proximal and distal joint and contralateral distal joint effects may occur with a unilateral knee dysfunction. This is in agreement with a previous study examining knee joint flexion restrictions (Howard et al. in revision) and a study evaluating squat performance with 8 subjects who had undergone ACL reconstruction

(Salem et al., 2003). Salem et al. reported the reconstructed knee flexion excursion was 3.5° less, hip excursion 2.0° less and ankle excursion 2.5° less when compared to the uninvolved limb in subjects with a mean length of time one year post operative.

Decreased flexion excursion may be cause for concern when using the squat to selectively target the lower extremity for rehabilitation or sports performance with populations who have recently undergone surgery or injury to the knee. This may result in asymmetrical joint excursion across the ipsilateral and contralateral hip, knee and ankle which may limit the overall effectiveness of the squat in stimulating multiple joints through a fully functional ROM. If the asymmetry is deemed clinically significant (being that the restriction causes significant compensation) the squat depth may need to be modified by the clinician until an optimal amount of symmetry is observed, thus resulting in a relative equal load between limbs.

The findings of the present study supports the hypothesis of knee joint restrictions creating lower extremity kinematic changes during the squat by identifying a potential compensatory mechanism of ipsilateral and contralateral limb substitution patterns existing in subjects squatting with a knee dysfunction. Although not a part of the current study, joint excursions in non-sagittal planes may also alter joint position centers and contribute to ipsilateral and contralateral limb compensation. Again, this study did not evaluate non sagittal biomechanics which may be most prevalent at the hip given the magnitude of transverse plane and frontal plane motion available at this joint.

Ankle Joint Specific Restrictions

Another purpose of this study was to evaluate the effect of a unilateral ankle dorsiflexion restriction. Unlike the knee it was felt a brace would not adequately restrict joint motion without excessively compromising squat stance position. The primary concerns were that a device applied unilaterally may alter the stance position and direct weight transfer onto the force plate. Therefore, an indirect ankle dorsiflexion restriction was created by unilaterally blocking the knee from advancing forward past the toes (Fry et al. 2003). The ipsilateral ankle restriction produced 16.2° ± 3.4° compared to the normal condition 22.7° ± 5.7°, yielding approximately 6° of ipsilateral ankle restriction resulting in a large effect (1.47). Again this produced a compensation visually mimicking what one may see in the clinic when observing an individual squatting with an ankle dorsiflexion dysfunction. Typical clinical conditions that may result in such dysfunction include ankle sprains, peroneal subluxation, achilles tendonitis, intraarticular fractures, and fusions.

The ankle restriction decreased ipsilateral ankle and knee motion but produced no significant changes in hip flexion ROM. Ipsilateral knee flexion and ankle dorsiflexion decreased 7° and 6° respectively to the non braced condition signifying a proximal effect on the distal ankle restriction. A possible explanation for this effect is that restricting tibial anterior displacement relative to the ankle results in a secondary knee flexion restriction due to the kinetic chain relationship with the two joint segments. During the squat, knee flexion and ankle dorsiflexion appear to be coupled movements, thus a restriction at the ankle will limit knee flexion. If the restriction occurs prior to achieving the desired squat depth the patient/subject must adjust his/her mechanics which may result in the overall COM shifting posterior and toward the contralateral limb compared to the non braced condition (Howard et al., in revision). Another conceivable adjustment is for the patient/subject to flex excessively at the hip to offset the lack of ankle dorsiflexion (anterior knee displacement) but since instructions were to maintain an upright posture during the squat, which is proper form, this potential compensation was likely controlled.

The ankle restriction decreased contralateral hip flexion 2.3° while contralateral ankle dorsiflexion decreased 2° compared to the non braced condition. Interestingly, there were no significant sagittal plane ROM changes in the contralateral knee which may be partially explained by proximal and distal joint changes relative to the knee in the sagittal and non sagittal planes thus eliminating the need for sagittal plane knee compensation. Furthermore, the contralateral ankle effects were small when compared to the ipsilateral limb and may not have been adequate to affect the contralateral knee.

These ankle restricted findings support the hypotheses comparing squatting kinematics across conditions. It appears that joint restrictions affect the ipsilateral limb by limiting proximal and distal joint excursion relative to the involved site. Similar effects are noted at the contralateral hip and ankle. All subjects were instructed to maintain a forward facing foot position to control the variability and potential kinematic effects that toe out angles may have on lower extremity (Ninos et al., 1997 & Escamilla et al., 2001). Future studies should explore the effect of joint restrictions with preferred toe out stance positions. Also, when subjects were at maximum descent during the squat, noticeable asymmetry at the lumbar paraspinals and lumbopelvic region was noted by the P.I. While the intent of this study was isolated to the hip, knee and ankle, one should not discount the role that the lumopelvic region may play in accommodating lower extremity joint dysfunctions.

Energetics

The second objective of this study was to compare sagittal plane energetics at the ankle, knee and hip during squatting in non braced, knee restricted and ankle joint dorsiflexion restricted conditions. The hypothesis of contralateral limb energetic demands being greater during the restricted conditions when compared to the ipsilateral limb was not supported. However, several interesting findings emerged regarding the transfer of work on limbs between conditions. All conditions and both limbs demonstrated that the most work was done on the knee joint followed by the hip and then ankle. The knee restriction decreased work at the ipsilateral knee, while the ankle restriction increased work at the ipsilateral knee and decreased work at the ipsilateral ankle. Moderate to large effects were noted between limbs at the hip, knee and ankle in the non braced condition suggesting limb dominance during "normal" squats. There were no significant contralateral joint changes between conditions.

Knee Joint Specific Restrictions

The reduction of work at the ipsilateral knee should be expected as the brace restricted approximately 13° of ipsilateral knee ROM excursion resulting in less work potential on the knee (as work is simply defined as the product of torque and angular displacement). Given the relationship of work to torque and angular displacement, comparison to previous literature addressing torque differences may offer insight to compensation patterns. Although not specifically addressing work, Salem et al. (2005)

reported a significantly decreased knee extensor peak moment (1.02 ± 0.31 Nm/kg) in ACL reconstructed knees compared to (1.28 ± 0.28 Nm/kg) the normal and contralateral knee during the squat. Likewise, subjects averaging 9.3 months post operative ACL reconstruction demonstrated significantly decreased knee extensor moments during vertical jump take off (1.05 Nm/kg), landing (1.30 Nm/kg) and lateral step-ups (0.98 Nm/kg) compared to matched controls of 1.43 Nm/kg, 1.91 Nm/kg, and 1.33 Nm/kg respectively (Ernst et al., 2000). Given these findings it would be expected that there would be a corresponding decrease in work.

The knee restricted condition produced a small effect (.40) of increased work performed on the ipsilateral ankle. Combined with the kinematic data which indicated decreased ipsilateral dorsiflexion (6°) and with work being the integral of torque and joint angular velocity or the product of torque and angular displacement it suggests that although there was a reduction in ROM, the joint was moving at a higher angular velocity with in turn may have resulted in greater loading across the joint. A previous study examining the effects of knee dysfunction in subjects performing a stair climbing activity 6 months post operative ACL reconstruction, reported ipsilateral knee work decreased by 0.25 Nm/kg whereas contralateral ankle joint work increased 0.09 Nm/kg compared to pre-operative values (Kowalk et al., 1997). Their findings suggested contralateral compensation in the kinetic chain. This is in contrast to the present study where there was no increased work reported at the contralateral limb, despite the ipsilateral reduction in knee and ankle work. These global compensation patterns will be addressed in the section following ankle specific restrictions.

Ankle Joint Specific Restriction

The ankle restriction produced a small effect (.22) of decreased work at the ipsilateral ankle. There was a trend toward decreased work on the ipsilateral knee (ES = .40) but this did not reach significance. There were no significant contralateral limb joint effects noted between conditions. These findings indicate that the ankle joint restriction primarily affected the restricted site and that neither the proximal joint segments nor the contralateral limb were affected. To date, the primary author has been unable to locate any comparable published scientific studies evaluating the effects of a unilateral ankle joint restriction during the squat.

An explanation as to why significant energetic changes did not occur in the contralateral limb during the ankle restricted condition is that the subjects in the current study may have compensated by limiting bilateral ankle dorsiflexion in an attempt to maintain limb symmetry. Another factor to consider is the inherent variability of ankle dorsiflexion used in squatting. Some subjects prefer to squat with a vertical tibial orientation, thus limiting ankle dorsiflexion, yet others maximize dorsiflexion resulting in increased tibial angulation relative to the foot (see Figures 10a & b). In this study the subjects who preferentially squatted with less ankle dorsiflexion may not have received sufficient unilateral ankle restriction potentially underestimating the biomechanical effects of an ankle restriction when squatting, thus obscuring any actual changes occurring.

Figure 10a. An example of the squat with knees anterior to the toes near the bottom of descent whereas 10b the knees are in line with the toes near the bottom of descent.

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Energetic Compensation Issues

While contralateral limb energetics did not significantly increase at the hip, knee or ankle, some interesting trends within both limbs occurred. Statistical analyses were not performed on the following observations. The summed total work (hip + knee + ankle) indicated that work done on the contralateral limb increased in the knee restricted condition, while work on the ipsilateral limb decreased. During the ankle restricted condition work on the ipsilateral limb decreased but there was no proportionate increase in work on the contralateral limb. Therefore future investigations of changes between and within limbs may be beneficial when studying the complex biomechanics of the normal and joint impaired squat. This method may help explain how work done on the joint shifts during joint restricted conditions, perhaps better capturing risk factors for reinjury or secondary injury. While the original hypotheses did evaluate changing hip, knee and ankle work between conditions, table 8 contains absolute values as well as the work percentage each joint contributed during the squat for the specific condition (calculated by summing the individual joint work per limb and dividing each respective joint by the limb sum and multiplying that number by 100).

Table 8. **Absolute work means and standard deviations normalized to bodyweight (Nm/kg); Contralateral (non-braced) and ipsilateral (braced) limb sagittal plane work contributions from the hip, knee and ankle (percentages).**

	Non braced Nm/kg	Knee Restricted Nm/kg	Ankle Restricted Nm/kg
Contralateral hip	0.13 ± .08 (22%)	0.14 ± .11 (23%)	0.11 ± .08 (19%)
Ipsilateral hip	**0.21 ± .19 (28%)**	**0.22 ± .18 (32%)**	**0.20 ± .20 (30%)**
Contralateral knee	0.43 ± .12 (74%)	0.45 ± .14 (75%)	0.45 ± .13 (79%)
Ipsilateral knee	**0.49 ± .23 (66%)**	**0.40 ± .20 (59%)**	**0.44 ± .21 (66%)**
Contralateral ankle	0.02 ± .02 (4%)	0.01 ± .01 (2%)	0.01 ± .01 (2%)
Ipsilateral ankle	**0.04 ± .05 (6%)**	**0.06 ± .05 (9%)**	**0.03 ± .04 (4%)**
Total work contralateral limb*	0.58 ± .09	0.60 ± .10	0.57 ± .09
Total work ipsilateral limb*	**0.74 ± .17**	**0.68 ± .15**	**0.67 ± .17**

*Total work calculated as the absolute hip, knee and ankle values summed

A closer look at the intralimb changes reveals some interesting findings. The knee restriction effectively had no influence on contralateral hip and knee contributions to summed relative work, while contralateral ankle relative work minimally decreased compared to the non braced condition. The ankle restriction seemingly had a larger contralateral effect as contralateral hip and ankle work contribution decreased while knee contribution increased. These findings suggest knee and ankle joint dysfunctions may result in slightly different joint energetic contralateral limb changes from "normal" during the squat.

The ipsilateral limb also revealed changes in the distribution of relative work between conditions. During the knee restricted condition relative work increased at the ipsilateral hip and ankle, while decreasing at the knee when compared to the non braced condition. The ankle restriction relative work increased at the ipsilateral hip, decreased at the ipsilateral ankle and produced no change at the knee when compared to the normal condition. These findings suggest that knee restrictions may have a greater effect on ipsilateral limb biomechanics compared to ankle dysfunctions. Additionally it suggests that a percentage of the total work is redistributed to the other non-restricted joints of the ipsilateral limb.

The current study is in agreement with previous literature confirming most work is performed on the knee during squatting (Escamilla et al., 2001,) and that knee dysfunction appears to decrease ipsilateral knee joint moments, while the ipsilateral hip and ankle compensate for this void (Kowalk et al., 1997, Ernst et al., 2000). These intra limb findings may be important in determining which joints receive inadequate stimulation or excessive stimulation, either of which could be deleterious when recovering from an injury.

In addition to intralimb findings, work percentage at the hip, knee and ankle were not symmetrical between limbs or across conditions in this study. The summed total work in the non braced condition on the ipsilateral limb was 0.74 (± .17) Nm/kg, compared to the contralateral limb where work was 0.58 (± .09) Nm/kg, netting a normalized to bodyweight limb difference of 0.16 Nm/kg. When comparing limbs across conditions the non braced interlimb differences account for some of the greatest disparity in kinematics and energetics especially across the hip and knee. Normal interlimb differences have been reported in previous work examining lower extremity peak joint moments derived from kinematic and GRF data that questioned the assumption of bilateral symmetry during a sit to stand movement (Lundin et al., 1995). The authors reported that assuming bilateral GRF symmetry underestimated peak moments at the ankles, knees and hips with the greatest disparity occurring at the hips ranging from 5.6 Nm to 15 Nm. Rodeosky et al. (1989) examined joint kinematic and moment symmetry during sit to stand and reported left to right asymmetries for ankle dorsiflexion, knee moment and hip moment. Although, neither of the authors reported changes in work across the joints/limbs they add to a growing notion that the clinician should not automatically assume interlimb symmetry, even in a “healthy” population.

The kinematic and energetic limb asymmetries reported in this study are interconnected. As previously mentioned, work is the product of torque and angular displacement, therefore the joint with less excursion will have less work associated with it unless the joint was moving at a higher angular velocity with may have result in a greater torque across the joint. The primary author has been unable to locate any published studies determining what constitutes normal or clinically acceptable symmetry during the squat, but feel in light of these findings further study may better define these parameters.

Squat performance is likely subject to wide ranges of individual variability due to the multiple joints and the respective degrees of freedom collectively involved in completing the task. This variability may explain how some subjects are better able to complete the task with smaller magnitudes of change. For example, in the current study most of the variability as defined by the standard deviations occurred at the hip, followed by the knee and then ankle. It is conceivable that subjects in the restricted conditions who were less efficient in shifting work demands within and between limb joints produced the greatest kinematic effects and may be at the greatest risk for primary or secondary injury. While subjects who were more efficient shifting work demands during knee restrictions were able to resolve the degrees of freedom restrictions with less of an effect. Clinical examples include post operative conditions like ACL reconstructions, meniscal arthroscopies and non operative knee conditions like patella tendonitis, knee sprains, contusions and patellofemoral pain syndrome. Length of post operative time, pain and weakness may also factor into the amount of compensation when squatting (Agre & Baxter, 1987; Salem et al., 2003). What remains unknown is the critical point at which this becomes problematic and if these effects are temporary or long term.

The squat is a reciprocal movement and previous work examining the squat with no external resistance (as was used in the current study) reports no significant changes in hip, knee and ankle joint powers between concentric and eccentric phases (Flannigan et al., 2003). Since this study controlled the cadence of the squat and no external resistance was applied to the exercise only the eccentric portion of the squat was analyzed.

Clinical Relevance of the Squat for Rehabilitation

Sports Performance and Injury Prevention

The squat continues to serve as a primary exercise for lower extremity rehabilitation and performance enhancement; however, this study's findings demonstrate potential concerns when squatting with existing ankle or knee dysfunction. The collective findings of this study should raise awareness of professionals working with populations known to have experienced significant injury that results in a relative long term loss of ROM. What remains unknown is the short and long term consequences of early return to activity prior to achieving "normal" joint biomechanics and if it could have the corresponding potential to lead to primary reinjury or secondary injury. What does seem clear is that the ipsilateral limb has the ability to shift work to proximal and distal joints from the dysfunctional site.

This has practical significance to clinicians as these substitutions in work could result in overuse (secondary) injury to the compensatory site or insufficient loading to the dysfunctional site, rendering it weak and susceptible to additional primary injury or limiting the athlete from achieving rehabilitation or performance goals. This scenario could exist in patients or athletes who have dysfunctions that are not overtly evident when performing squats or other functional tasks (Salem et al., 2003). If common patterns of compensations are known, clinicians can address the pertinent issues when designing rehabilitation programs. Most of the compensations in the current study occurred in the ipsilateral limb suggesting effects from the joint dysfunction occur proximal and distal to the involved joint. Since coronal and transverse planes were not examined in this study it is difficult to conclude how these planes may be affected by joint dysfunction. It is a possibility that a secondary injury mechanism may exist that is the result of either an overloading or underloading of the joints adjacent to the joint of primary dysfunction.

There is evidence that patients who are post-operative ACL reconstruction perform stair climbing with less work at the knee and more work at the hip and ankle compared to the contralateral limb (Kowalk et al., 1997). Interestingly, when comparing total work (hip + knee + ankle) differences between limbs were minimal (Kowalk et al., 1997). This would indicate the limb was able to effectively shift work to the proximal and distal joint to maintain total limb symmetry. It remains unknown if this most dominantly places the proximal or distal compensatory sites at risk of a secondary injury or reinjury to the primary site.

The current study suggests total work between limbs appears asymmetrical during the squat (Table 8). It is important to note that the squat ROM used in this study required approximately 50 degrees more hip flexion and 25 degrees more knee flexion compared to Kowalk et al. (1997), thus work potential and compensation would appear greater due to larger joint excursion. The amount of limb asymmetry in the non braced condition warrants further study. This may simply represent the independence of the limbs to function based upon the daily demands placed on the body and nothing more than a normative level found in the population studied.

Clinicians often benchmark the integrity of the athlete's injury to the contralateral site but if the comparison joint happens to be a part of the weak link in the kinetic chain, the comparison may be invalid. Moreover, if the injured limb predominately relied on the dysfunctional site's (pre injury) energetics to accomplish daily and sporting activities at a greater percentage than the adjacent joints or the contralateral comparison, a potential concern could be reinjury or performance deficits when returning to sport. This makes it essential for the clinician to have more than one assessment tool for proximal and distal comparisons to be included in the evaluation.

The concept of isolated joint dysfunction causing or being caused by risk factors such as weakness or pain elsewhere in the lower extremity has been a focus of previous work (Bullock-Saxon, 1994). Hip extensor neuromuscular deficits were reported in subjects with a history of severe ankle injuries performing prone hip extension (Bullock-Saxon et al.1994). The primary limitation of this finding is the inability to determine whether the injury caused the deficit or the deficit was the result of the injury. In the current study the ankle restricted condition did not produce significant hip energetic changes between conditions but did result in shifting a percentage of total contralateral work on the hip and ankle to the knee. While relative work increased at the ipsilateral hip and decreased at the ipsilateral ankle, it resulted in no relative change of work contribution on the knee (see Tables 11 & 12). Relative work contributions can then be compared to the absolute joint values (see Tables 11 & 12). This is important because it is conceivable there could be no change in relative work contribution from the individual joints but an overall increase or decrease in total work. In the current study the ankle restriction caused a decrease and shift in total work (hip + knee +ankle) on the ipsilateral limb compared to the non braced condition. The contralateral limb showed little net change suggesting work done on the joints shifted to other areas that were unaccounted for in this study. The most likely region is the lumbar spine as it is the nearest major joint complex to the hips. This suggests a possible link between primary and secondary dysfunctions and highlights the importance of a thorough clinical evaluation that includes proximal and distal screening to the injured site.

Another clinical concern centers on work absorption changes across the lower extremities and lumbopelvic complex. Empirically, asymmetry can occur at the lumbopelvic hip complex when squatting with a joint restriction at the knee or ankle. During the descent phase of the squat, the hip, knee and ankle attenuate ground reaction forces through negative mechanical work. However, work done on the lumbar spine was not assessed in this study. The trunk is often portrayed in biomechanical modeling as a rigid segment, when in reality work is done at various spinal segments that may not be adequately measured through hip absorption (Kulas dissertation, 2005). This may have resulted in omission of lumbopelvic contributions that could potentially explain a portion of the compensations that occurred in the joint restricted conditions. Kingma et al. (2004) reported L4/5 spinal shear forces of 300N and L5/S1 shear forces ranging from 1100 – 1400 N when squatting with 10.5 kg of resistance. If the lumbo pelvic work absorption values were also known, they may likely show the lumbar spine as a key contributor when squatting (Lander et al., 1986). The primary author has found no studies examining the contribution of the lumbar spine when squatting with a lower extremity joint dysfunction.

Limitations

The primary limitation to this study was that rather than using "injured" subjects an artificial joint restriction at the knee and ankle was created to examine the effects of joint restrictions on squatting. Previous reliability testing demonstrated that subjects were able to perform a joint restricted squat ($ICC_{2,k}$ = 0.63-0.88) with equal consistency as that found in the normal squat ($ICC_{2,k}$ = 0.62-0.82), thus supporting a mechanical restriction as a reliable model for simulating and investigating biomechanical effects resulting from range of motion restrictions (Howard et al., in revision). Although reliable, one could question the model's external validity in patients with knee and ankle joint dysfunctions. It is important to note this study's findings of decreased ipsilateral ankle and knee kinematics and decreased knee kinetics are similar to previous studies examining the squat with patients who are recovering from knee injuries (Neitzel et al., 2002; Salem et al., 2003). The current study's findings use the non braced condition as a control, where others use the contralateral limb as the control. Comparing the current study's findings with Salem et al., ipsilateral knee kinematics decreased 6° versus 3°, whereas ipsilateral ankle ROM decreased 5° versus 2.5°. Therefore, this study may best serve as a general sagittal plane model for clinicians and coaches to reference when using the parallel squat in patients/athletes with knee and ankle dysfunction.

Another limitation of this study is that compensations due to injury may be mediated by altered neuromuscular strategies and the training effects produced by rehabilitation protocols (Devita et al., 1996). These changes may not be taken into account using a mechanical device to create a joint restriction, therefore surface EMG may be beneficial in assessing neuromuscular changes between conditions.

Future Studies

The results of this dissertation indicate that a joint restriction at the knee or the ankle produces sagittal plane biomechanical changes in the lower extremities. In order to further support and explain the current findings, transverse and coronal plane hip and knee kinematics and kinetics during the squat are necessary. Previous work has demonstrated transverse and coronal plane hip and knee ipsilateral and contralateral compensations during a knee restricted squat (Howard et al., in revision). Unfortunately, these findings were unable to be reported for the current study due to technical malfunction. Although the squat exercise is considered a sagittal plane dominant exercise, this information would better clarify lower extremity compensations during a joint restriction.

A prospective study tracking healthy subjects who regularly engage in squatting exercises would allow the researcher to track lower extremity injuries and examine the short and long term changes in lower extremity biomechanics when squatting. Additionally, surface EMG of lower extremity and trunk musculature would be helpful by describing changes in muscle activation patterns. Combining EMG with joint power and ROM values would provide the clinician/coach with an unparalleled understanding of the effects of joint dysfunctions when performing squats. Clinicians could use this information to develop screening tools and treatment strategies to correct lower extremity faulty movement patterns.

While the current study standardized foot position and stance width, it may be beneficial to examine how subjects self selected stance widths and foot positions influence lower extremity biomechanics and potentially change as a result of injury. This information may enhance identification of faulty movement patterns and assist the professional in "customizing" squat stance and foot position for increased efficacy.

Conclusions

This study demonstrated that isolated joint restrictions at the ankle and knee produced compensatory changes in normal lower extremity biomechanics when squatting. In the ankle restricted condition, ipsilateral ankle, knee and hip sagittal plane ROM was decreased while contralateral ankle and hip sagittal plane ROM also decreased. There were no significant sagittal plane work changes in either limb with the ankle restriction. The knee restricted condition produced decreased sagittal plane ipsilateral ankle, knee and hip ROM, while no significant kinematic changes occurred in the contralateral limb. There was decreased work done on the ispsilateral knee and increased work done on the ipsilateral ankle with a trend toward changes in the relative intralimb ankle, knee and hip work compared to the non braced condition.

The results of this study may best be viewed as a beginning model depicting biomechanical compensations that can occur when squatting with a joint dysfunction. Future research is needed in healthy subjects to examine whether faulty movement patterns occur in the sagittal, coronal and transverse planes at the hip, knee and ankle in response to injury and how long these changes last. Additionally, lumbopelvic biomechanics should be included in the analysis. This information may prove to be beneficial in developing pre-participation screening tools, treatment strategies and identifying risk factors for secondary injury.

BIBLIOGRAPHY

AGRE, J.C., BAXTER, T.L. Musculoskeletal profile of male collegiate soccer players. *Arch. Phys. Med. Rehabil*. 68:147-150. 1987.

ALLARD, P.I., A.F.STOKES, and J.P. BLANCHI. *Three dimensional analysis of human movement*. Champaign,IL: Human Kinetics, 1995.

ARANGIO, G.A., C. CHEN, M. KALADY, and JF REED. Thigh muscle size and strength after anterior cruciate ligament reconstruction and rehabilitation. *J. Orthop. Sports Phys. Ther.* 26: 238-43. 1997.

ATHA, J. Strengthening muscle. *Exercise and sports science reviews*. 7: 163-172. 1981.

AUGUSTSSON, J., A. ESKO, R. THOMEE, and U. SVANTESSON. Weight training of the thigh muscles using closed vs. open kinetic chain exercises: a comparison of performance enhancement. *J. Orthop. Sports Phys. Ther.* 27: 3-8. 1998.

BEYNNON, B.D., R.J. JOHNSON, B.C. FLEMMING, C.J. STANKEWICH, P.A. RENSTROM, and C.E. NICHOLS. The strain behavior of the anterior cruciate ligament during squatting and active flexion-extension: a comparison of an open and closed kinetic chain exercise. *Am. J. Sports Med* 25: 823-829. 1997.

BLACKBURN, J.T., B.L. RIEMANN, J.B.MYERS and S.M. LEPHART. Kinematic analysis of the hip and trunk during bilateral stance on firm, foam, and multiaxial support surfaces. *Clinical Biomechanics*.18: 655-66. 2003.

BULLOCK-SAXTON, J E; JANDA, V; BULLOCK, M I. The influence of ankle sprain injury on muscle activation during hip extension. *International Journal Of Sports Medicine*. 15: 330-334. 1994.

BUTLER, R. J., CROWELL, H. P., and I. MCCLAY DAVIS. Lower extremity stiffness: implications for performance and injury. *Clinical Biomechanics*. 18: 511-517. 2003.

BYNUM, B.E., R.L. BARRACK, and A.H. ALEXANDER. Open versus closed chain kinetic exercises after anterior cruciate ligament reconstruction. *Am. J. Sports Med.* 23: 401–406. 1995.

CHANFLER, T.J., G.D. WILSON, and M.H. STONE. The effect of the squat exercise on knee stability. *Med. Sci. Sports Exerc*. 21: 299303. 1989.

CHANDLER, T.J. and M STONE. The squat exercise in athletic conditioning: A position statement and review of the literature. *National Strength and Conditioning Journal*. 13: 51-58. 1991.

COQUEIRO, K.R. D. BEVILAQUA-GROSSI, F. BERZIN, A.B.SOARES, C. CANDOLO and V. MONTEIRO-PEDRO. Analysis on the activation of the VMO and VLL muscles during semisquat exercises with and without hip adduction in individuals with patellofemoral pain syndrome. J *Electromyogr Kinesiol*. 15: 596-603. 2005.

DEMPSTER, W. T. Space requirements of the seated operator *(WADC Technical Report).* Wright-Patterson Air Force Base, Ohio. 1955.

DEVITA, P. BLANKENSHIP, H.P.and W. SKELLY. Effects of a functional knee brace on the biomechanics of running. *Medicine and Science in Sports and Exercise*. 24: 797-806.1992a.

DEVITA, P. AND W.A. SKELLY. Effect of landing stiffness on joint kinetics and energetics in the lower extremity. *Medicine and Science in Sports and Exercise*. 24: 108-115. 1992b.

DEVITA, P., M. TORRY, K.L. GLOVER AND D.L. SPERONI. A functional knee brace alters joint torque and power patterns during walking and running. *J.*

DUNN, B., K. KLEIN, B. KROLL. T. MCLAUGHUN, P. O'SHEA, and D. WATHEN. Coaches round table: The squat and its application to athletic performance. *Strength Cond. J.* 6:10-22. 1984.

EKSTRAND, J. and J. GILLQUIST. Soccer injuries and their mechanisms: a prospective study. *Med. Sci. Sports Exerc.* 15: 267-270. 1983.

ESCAMILLA, R.F., G.S. FLEISIG, N. ZHENG, S.W. BARRENTINE, K.E. WILK, and J.R. ANDREWS. Biomechanics of the knee during closed kinetic chain and open kinetic chain exercises. *Med Sci Sports Exerc* 30: 556–569.1998.

ESCAMILLA, R. F., A.C. FRANCISCO, and G. S. FLEISIG. A three-dimensional biomechanical analysis of sumo and conventional style deadlifts. *Med. Sci. Sports Exerc*. 32: 1265–1275. 2000.

ESCAMILLA, R. F., N. ZHENG, and G. S. FLEISIG. The effects of technique variations on knee biomechanics during the squat and leg press. *Med. Sci. Sports Exerc.* 29: S156. 2001.

ESCAMILLA, RF., G.S. FLEISIG, T.M LOWRY, S.W. BARRENTINE, and J.R. ANDREWS. A three-dimensional biomechanical analysis of the squat during varying stance widths. *Med. Sci. in Sports & Ex.* 33: 984-998. 2001.

ENG, J.J., and D.A. WINTER. Kinetic analysis of the lower limbs during walking: What information can be gained from a threee-dimensional model? *J. Biomechanics*. 28: 453-758. 1995.

ERNST, G.P., E. SALIBA, D.R. DIDUCH, S.R. HURWITZ and D.W. BALL. Lower-Extremity Compensations Following Anterior Cruciate Ligament Reconstruction. *Phys. Ther.* 80: 251-260. 2000.

FITZGERALD, G.K. Open versus closed kinetic chain exercise: Issues in rehabilitation after anterior cruciate ligament reconstructive surgery. *Phys. Ther.* 77: 1747–1754. 1997.

FLANAGAN, S., G.J.SALEM, M-Y WANG, S.E. SANKER, and G.A. GREENDALE. Squatting exercises in older adults: Kinematic and kinetic comparisons. *Med. & Sci. in Sports & Ex.* 35: 635-643. 2003.

FLECK, S.J., and W.J. KRAEMER. *Designing resistance training programs*. Champaign, Ill: Human Kinetics. 1987.

FRY, A.C., J. CHADWICK SMITH, and B.K. SCHILLING. Effect of knee position on hip and knee during the barbell squat. *Journal of Strength and Conditioning Research.* 17: 629-633. 2004

FU, F.H., S.L.-Y. WOO, and J.J. IRRGANG. Current concepts for rehabilitation following anterior cruciate ligament reconstruction. *J. Orthop. Sports Phys. Ther*. 15: 270-278. 1992..

GREENFIELD, B. *Functional Anatomy of the Knee. In: Rehabilitation of the Knee: A Problem Solving Approach*. B Greenfield, ed. Philadelphia, Davis. 1993.

HAMILL, J. and K.M. KNUTZEN. *Biomechanical Basis of Human Movement 2nd ed.* Baltimore, Lippincott Williams & Wilkins. 2003.

HERRICK, R.T., M.H. STONE, and S. HERICK. Injuries in strength –power activities. *Powerlifting USA*. 7: 7-9. 1983.

HOUCK, J. H.J. YACK, and T. CUDDEFORD. Validity and comparisons of tibiofemoral orientations and displacement using a femoral tracking device during early to mid stance of walking. *Gait and Posture.* 19: 76-84. 2004.

HOWARD, R.L. R.J. SCHMITZ, S.J. SHULTZ, K..WILLIAMS, D.H. PERRIN AND R. M.LUECHT. *Biomechanical Effects of a Sagittal Plane Joint Restriction during Squatting. Journal of Strength and Conditioning Research.* In Revision.

INMAN, W.S., J.M. TIMONEY, P.M. QUESADA, P.F. SHARKEY, H.B. SKINNER, R.L. BARRACK, and A.H. Alexander. The effect of anterior cruciate ligament reconstruction on gait. *Contemp. Orthop.* 30: 209-212, 1995.

KLEIN, K.K. The deep squat exercise as utilized in weight training for athletes and its effects on ligaments of the knee. *Journal of the Association for Physical and Mental Rehabilitation*. 15: 6-11. 1961.

KLEIN, K.K. An instrument for testing the medial and lateral collateral ligament stability of the knee. *Am, J. Surg*. 104: 798-772. 1962.

KOWALK, D.L., J.A. DUNCAN, F.C. McCUE III, and C.L. VAUGHAN. Anterior cruciate ligament reconstruction and joint dynamics during stair climbing. *Med. Sci. Sports Exerc*. 29: 1406-1413. 1997.

LANDER, J. E., B. T. BATES, and P. DEVITA. Biomechanics of the squat exercise using a modified center of mass bar. *Med. Sci. Sports Exerc*. 18: 469–478, 1986.

LOPRESTI, C., DT KIRKENDALL, G.M STREET, and A.W. DUDLEY. Quadriceps insufficiency following repair of the anterior cruciate ligament. *J. Orthop. Sports Phys. Ther.* 9: 245–49. 1988.

LUTZ, G.E., R.A. PALMITIER, K.N. AN, and E.Y.S. CHAO. Comparison of tibiofemoral joint forces during open kinetic chain exercises. *J. Bone Joint Surg.* 75A: 732–739. 1993.

MAZUR, L.J., R.J. YETMAN, and W.L. RISSER. Weight training injuries. Common injuries and preventative methods. *Sports Med*. 16 (1): 57-63. 1993.

MCBRYDE JR, A.M., and HOFFMAN, J.L. Injuries to the foot and ankle in athletes. *Southern Medical Journal*. 97: 738-741. 2004.

MCCAW, S. T., and D. R. MELROSE. Stance width and bar load effects on leg muscle activity during the parallel squat. *Med. Sci. Sports Exerc*. 31: 428–436. 1999.

MCLAUGHLIN, T. M., C. J. DILLMAN, and T. J. LARDNER. A kinematic model of performance in the parallel squat by champion powerlifters. *Med. Sci. Sports Exerc*. 9: 128–133. 1977.

MEYERS, E.J. Effect of selected exercise variables on ligament stability and flexibility of the knee. *Research Quarterly*. 42: 411-422. 1971.

NADLER, S.F., K.D. W.U., T. GALSKI and J.H. FEINBERG. Low back pain in college athletes. A prospective study correlating lower extremity overuse or acquired ligamentous laxity with low back pain. *Spine*. 23: 828-833.1998.

NEITZEL, J.A., T.W. KERNOZEK, and G.J. DAVIES. Loading response following anterior cruciate ligament reconstruction during the parallel squat exercise. *Clinical Biomechanics.* 17: 551-554. 2002.

NICHOLAS, J.A., R.B. GROSSMAN and E.B. HERSHMAN. The importance of a simplified classification of motion in sports in relation to performance. *Orthop. Clin. N. Am. 8: 499 -532. 1977.*

NINOS, J., J. IRRGANG, R. BURDETT, and J. WEISS. Electromyographic analysis of the squat performed in self selected lower extremity neutral rotation and 30 degrees of lower extremity turn-out from the self selected neutral position. *J. Orthop. Sports Phys. Ther.* 25:307-321. 1997.

NISELL, R., and J. EKHOLM. Joint load during the parallel squat in powerlifting and force analysis of in vivo bilateral quadriceps tendon rupture. *Scand J. Sports Sci.* 8: 63–70. 1986.

NORKIN, C.C., and P.K. LEVANGIE. *Joint Structure and Function 2nd ed.* Philadelphia, F.A. Davis Co. *1992.*

O'SHEA., P. The Parallel Squat. *Natl. Strength Cond.* J. 7:4-7.1985.

PALMITIER, R.A., A. KAI-NAN, S.G. SCOTT and E.Y.S. CHAO. Kinetic chain exercise in knee rehabilitation. Sports Medicine. 11: 402-413. 1991.

PANARIELLO, R.A., S.I. BACKUS, and J.W. PARKER. The effect of the squat exercise on anterior-posterior knee translation in professional football players. *Am. J. Sports Med*. 122:768-773. 1994.

PERIE, D.A., J. TATE, P. L. CHENG and G. A. DUMAS. Evaluation and calibration of an electromagnetic tracking device for biomechanical analysis of lifting tasks. *Journal of Biomechanics* 35: 293-297. 2002.

PETRELLA, J.K., J KIM, S.C. TUGGLE, S.R. HALL and M.M. BAMMAN. Age differences in knee extension power, contractile velocity and fatigability. *J. Appl Physiol*. 98: 211-220. 2005.

ROBERTSON, G.E., G.E. CALDWELL, J. HAMILL, G. KAMEN and S.N. WHITTLEDEY. *Research Methods in Biomechanics*. Champaigne, Ill. Human Kinetics. 2004.

ROBINIVICH, S.N., R. BRUMER, and J MAURER. Effect of the "squat response" on impact velocity during backward falls. *Journal of Biomechanics*. 37:1329-1337. 2004.

RUBENSTEIN, R.A., K.D. SHELBOURNE, C.D. VANMETER, J.C. McCARROLL, and A.C. RETTIG. Isolated autogenous bone-patellar tendon-bone graft site morbidity. *Am. J. Sports Med.* 22:324-327. 1994.

RUSSELL, P.J., and S.J. PJILLIPS. A preliminary comparison of front and back squat exercise. *Res. Q. Exerc. Sport.* 60: 201-208. 1989.

SALEM, G.R., R. SALINAS, V. HARDING. Bilateral kinematic and kinetic analysis of the squat exercise after anterior cruciate ligament reconstruction. Arch Phys Med Rehabil. 84: 1211-1216. 2003.

SANTOS, M.J., K. MCINTIRE, J. FOECKING, and D.W. LIU. The effects of ankle bracing on motion of the knee and the hip joint during trunk rotation tasks. *Clin. Biomech.* 19: 964-971. 2004.

SCHOT, P.K., K.M. KNUTZEN, S.M. POOLE, and L.A. MROTEK. Sit to stand performances of older adults following strength training. *Research Quarterly Exercise and Sport.* 74:1-8. 2003.

SHELBOURNE, K.D., and P. NITZ. Accelerated rehabilitation after anterior cruciate ligament reconstruction. *Am. J. Sports Med.* 18: 292–299. 1990.

SIGNORILE, J., J. KWIATKOWSKI, J. CARUSO, B. ROBERTSON, R. WILLIAMS, I. LOWENSTEYN, S. DIGEL, J. CARUSO, and W.G. LEBLANC. Effect of foot position on the electromyographical activity of the superficial quadriceps muscles during the parallel squat and knee extension. *J. Strength. Cond. Res.* 9: 182–187. 1995.

SODERMAN, K., H. ALFREDSON, and T. PIETILA, et al. Risk factors for leg injuries in f emale soccer players: a prospective investigation during one out-door season. *Knee Surg Sports Traumatol Arthrosc*. 9: 313-321. 2001.

STEIN, H.A., T. BROSKY, M.F. REINKING, J NYLAND, and M.B. MASON. Comparison of closed kinetic chain and isokinetic joint isolation exercise in patients with patellofemoral dysfunction. *J. Orthop. Sports Phys. Ther*. 24: 136-141. 1996.

STONE, M.H., R. BYRD, J. TEW and M. WOOD. Relationship between anaerobic power and olympic weightlifting performance. *Journal of Sports Medicine and Physical Fitness*. 20: 99-102. 1980.

STONE, M.H. Implicatons for connective tissue and bone alterations resulting from resistance exercise training. Med. Science Sports Exerc. 20: S162-S168. 1988.

STUART, M. J., D.A. MEGLAN, G. E. LUTZ, E. S. GROWNEY, and K.N. AN. Comparison of intersegmental tibiofemoral joint forces and muscle activity during various closed kinetic chain exercises. *Am. J. Sports Med*. 24: 792–799. 1996.

THOMEE, R., J. AUGUSTSSON, and J. KARLSSON. Patellofemoral pain syndrome: a review of current issues. *Sports Med*. 28: 245-62. 1999. Review.

TODD, T. Karl Klein and the squat. *NSCA Journal*. 3:26-31. 1984.

TOUTOUNGI, D.E., T.W. LU, A. LEARDINI, F. CATANI, and J.J. O'CONNER. Cruciate ligament forces in the human knee during rehabilitation exercises. *Clin Biomech* 15: 176-187. 2000.

TIBERIO, D. The effect of excessive subtalarjoint pronation on patellofemoral mechanics: A theoretical model. *J. Orthop. Sports Phys. Ther*. 9:160-165.1987.

TIBONE, J.E., and T.J. ANTICH. A biomechanical analysis of anterior cruciate ligament reconstruction witht the patella tendon. *Am. J. Sports Med.* 16:332-335. 1988.

WILK, K.E., R.F. ESCAMILLA, G.S. FLEISIG, S.W. BARRENTINE, J.R. ANDREWS and M.L. BOYD. A comparison of tibiofemoral joint forces and electromyographic activity during open and closed kinetic chain exercises. *Am. J. Sports Med.* 24:518-527. 1996.

WINTER, D. *Biomechanics and motor control of human movement*. New York: Wiley. 1990.

WORRELL, T.M., B. BORCHERT, K. ERNER, J. FRITZ and P. LEERAR. Effect of lateral step-up exercise protocol on quadriceps and lower extremity performance. *J Orthop Sports Phys Ther***.** 6: 646–653. 1993.

WRETENBERG, P., Y. FENG, F. LINDBERG, and U. P. ARBORELIUS. Joint moments of force and quadriceps activity during squatting exercise. *Scand. J. Med. Sci. Sports.* 3: 244–250. 1993.

WRETENBERG, P., Y. FENG, and U. P. ARBORELIUS. High-and low-bar squatting techniques during weight-training. *Med. Sci. Sports Exerc*. 28: 218–224. 1996.

YACK, H.J., L.A. WASHCO, and T.WHIELDON. Compressive forces as a limiting factor of anterior tibial translation in the ACL deficient knee. *Clin. J. Sports Med*. 4:233-239. 1994.

ZHANG, S. N., BATES, B. T., and J.S. DUFEK. Contribuations of lower extremity joints to energy dissipation during landings. *Med. Sci. Sports Exerc*. 32: 812-819. 2000.

Appendix A: Descriptive statistics of contralateral (left) and ipsilateral (right) hip and knee coronal and transverse plane kinematics across 3 conditions: 1) Normal, 2) Knee Restricted and 3) Ankle Restricted

Descriptive Statistics

	N	Mean	Std. Deviation
RHR1	42	10.1984	9.71342
RHA1	42	24.7780	8.73043
RKR1	42	-12.7217	10.56419
RKA1	42	1.3433	12.06814
LHR1	42	-28.0998	8.57757
LHA1	42	26.2431	11.89179
LKR1	42	-22.8387	13.08638
LKA1	42	-25.5973	12.21065
RHR2	42	3.8578	11.47075
RHA2	42	25.2354	7.61688
RKR2	42	-13.4081	7.98033
RKA2	42	-2.9631	10.01545
LHR2	42	-28.1742	10.64047
LHA2	42	30.0627	10.74768
LKR2	42	-19.7548	10.84327
LKA2	42	-23.3876	11.13242
RHR3	42	6.8629	13.17277
RHA3	42	26.1439	11.15083
RKR3	42	-11.4736	10.27387
RKA3	42	-2.1139	11.13461
LHR3	42	-29.2315	9.71339
LHA3	42	31.1033	13.21352
LKR3	42	-21.6051	10.55760
LKA3	42	-24.3799	12.17182

KEY: RHR: Right hip rotation (negative value indicates external rotation)
RHA: Right hip abduction (negative value indicates abduction)
RKR: Right knee rotation (negative value indicates external rotation)
RKA: Right knee abduction (negative value indicates abduction)
LHR: Left hip rotation (negative value indicates internal rotation)
LHA: Left hip abduction (negative value indicates adduction)
LKR: Left knee rotation (negative value indicates internal rotation)
LKA: Left knee abduction (negative value indicates adduction)

Appendix B: Descriptive statistics of contralateral (left) and ipsilateral (right) hip, knee and ankle sagittal plane kinematics across 3 conditions: 1) Normal, 2) Knee Restricted and 3) Ankle Restricted

Left and Right Hip Joint Displacement

	Mean	Std. Deviation	N
C1LH	103.6879	13.19471	42
C1RH	113.2488	11.85726	42
C2LH	103.0715	12.58300	42
C2RH	110.8423	13.22963	42
C3LH	101.3582	13.24736	42
C3RH	112.6510	13.36770	42

Left and Right Knee Joint Displacement

	Mean	Std. Deviation	N
C1LK	98.6787	8.93925	42
C1RK	106.8323	8.80441	42
C2LK	98.9883	8.90951	42
C2RK	93.0805	6.84220	42
C3LK	97.9303	8.12333	42
C3RK	99.7589	8.95458	42

Left and Right Ankle Joint Displacement

	Mean	Std. Deviation	N
C1LA	22.5001	5.65203	42
C1RA	22.7370	5.67543	42
C2LA	21.3306	5.78938	42
C2RA	17.4400	5.57939	42
C3LA	20.6643	5.20062	42
C3RA	16.1757	3.38528	42

Appendix C: Kinematics

Hip Kinematics General Linear Model: Repeated Measures (Condition x Limb)

Within-Subjects Factors

Measure: MEASURE_1

COND	LIMB	Dependent Variable
1	1	C1LH
	2	C1RH
2	1	C2LH
	2	C2RH
3	1	C3LH
	2	C3RH

Descriptive Statistics

	Mean	Std. Deviation	N
C1LH	103.6879	13.19471	42
C1RH	113.2488	11.85726	42
C2LH	103.0715	12.58300	42
C2RH	110.8423	13.22963	42
C3LH	101.3582	13.24736	42
C3RH	112.6510	13.36770	42

Multivariate Tests[b]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
COND	Pillai's Trace	.056	1.182[a]	2.000	40.000	.317	.056
	Wilks' Lambda	.944	1.182[a]	2.000	40.000	.317	.056
	Hotelling's Trace	.059	1.182[a]	2.000	40.000	.317	.056
	Roy's Largest Root	.059	1.182[a]	2.000	40.000	.317	.056
LIMB	Pillai's Trace	.503	41.439[a]	1.000	41.000	.000	.503
	Wilks' Lambda	.497	41.439[a]	1.000	41.000	.000	.503
	Hotelling's Trace	1.011	41.439[a]	1.000	41.000	.000	.503
	Roy's Largest Root	1.011	41.439[a]	1.000	41.000	.000	.503
COND * LIMB	Pillai's Trace	.186	4.570[a]	2.000	40.000	.016	.186
	Wilks' Lambda	.814	4.570[a]	2.000	40.000	.016	.186
	Hotelling's Trace	.228	4.570[a]	2.000	40.000	.016	.186
	Roy's Largest Root	.228	4.570[a]	2.000	40.000	.016	.186

a. Exact statistic

b.

Design: Intercept
Within Subjects Design: COND+LIMB+COND*LIMB

Mauchly's Test of Sphericity[b]

Measure: MEASURE_1

					Epsilon[a]		
Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Greenhouse-Geisser	Huynh-Feldt	Lower-bound
COND	.831	7.423	2	.024	.855	.889	.500
LIMB	1.000	.000	0	.	1.000	1.000	1.000
COND * LIMB	.694	14.636	2	.001	.765	.789	.500

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in Tests of Within-Subjects Effects table.

b.

Design: Intercept
Within Subjects Design: COND+LIMB+COND*LIMB

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
COND	Sphericity Assumed	124.016	2	62.008	1.221	.300	.029
	Greenhouse-Geisser	124.016	1.710	72.510	1.221	.296	.029
	Huynh-Feldt	124.016	1.777	69.774	1.221	.298	.029
	Lower-bound	124.016	1.000	124.016	1.221	.276	.029
Error(COND)	Sphericity Assumed	4164.724	82	50.789			
	Greenhouse-Geisser	4164.724	70.123	59.392			
	Huynh-Feldt	4164.724	72.873	57.150			
	Lower-bound	4164.724	41.000	101.579			
LIMB	Sphericity Assumed	5735.533	1	5735.533	41.439	.000	.503
	Greenhouse-Geisser	5735.533	1.000	5735.533	41.439	.000	.503
	Huynh-Feldt	5735.533	1.000	5735.533	41.439	.000	.503
	Lower-bound	5735.533	1.000	5735.533	41.439	.000	.503
Error(LIMB)	Sphericity Assumed	5674.822	41	138.410			
	Greenhouse-Geisser	5674.822	41.000	138.410			
	Huynh-Feldt	5674.822	41.000	138.410			
	Lower-bound	5674.822	41.000	138.410			
COND * LIMB	Sphericity Assumed	130.261	2	65.130	7.082	.001	.147
	Greenhouse-Geisser	130.261	1.531	85.088	7.082	.004	.147
	Huynh-Feldt	130.261	1.578	82.528	7.082	.003	.147
	Lower-bound	130.261	1.000	130.261	7.082	.011	.147
Error(COND*LIMB)	Sphericity Assumed	754.077	82	9.196			
	Greenhouse-Geisser	754.077	62.767	12.014			
	Huynh-Feldt	754.077	64.714	11.652			
	Lower-bound	754.077	41.000	18.392			

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	COND	LIMB	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
COND	Linear		89.984	1	89.984	2.371	.131	.055
	Quadratic		34.032	1	34.032	.535	.469	.013
Error(COND)	Linear		1556.217	41	37.957			
	Quadratic		2608.507	41	63.622			
LIMB		Linear	5735.533	1	5735.533	41.439	.000	.503
Error(LIMB)		Linear	5674.822	41	138.410			
COND * LIMB	Linear	Linear	31.493	1	31.493	5.513	.024	.119
	Quadratic	Linear	98.768	1	98.768	7.790	.008	.160
Error(COND*LIMB)	Linear	Linear	234.217	41	5.713			
	Quadratic	Linear	519.860	41	12.680			

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Intercept	2910908.481	1	2910908.481	3913.264	.000	.990
Error	30498.131	41	743.857			

Appendix D:

Knee Kinematics General Linear Model: Repeated Measures (Condition x Limb)

Within-Subjects Factors

Measure: MEASURE_1

COND	LIMB	Dependent Variable
1	1	C1LK
	2	C1RK
2	1	C2LK
	2	C2RK
3	1	C3LK
	2	C3RK

Descriptive Statistics

	Mean	Std. Deviation	N
C1LK	98.6787	8.93925	42
C1RK	106.8323	8.80441	42
C2LK	98.9883	8.90951	42
C2RK	93.0805	6.84220	42
C3LK	97.9303	8.12333	42
C3RK	99.7589	8.95458	42

Multivariate Tests[b]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
COND	Pillai's Trace	.698	46.199[a]	2.000	40.000	.000	.698
	Wilks' Lambda	.302	46.199[a]	2.000	40.000	.000	.698
	Hotelling's Trace	2.310	46.199[a]	2.000	40.000	.000	.698
	Roy's Largest Root	2.310	46.199[a]	2.000	40.000	.000	.698
LIMB	Pillai's Trace	.022	.903[a]	1.000	41.000	.347	.022
	Wilks' Lambda	.978	.903[a]	1.000	41.000	.347	.022
	Hotelling's Trace	.022	.903[a]	1.000	41.000	.347	.022
	Roy's Largest Root	.022	.903[a]	1.000	41.000	.347	.022
COND * LIMB	Pillai's Trace	.741	57.245[a]	2.000	40.000	.000	.741
	Wilks' Lambda	.259	57.245[a]	2.000	40.000	.000	.741
	Hotelling's Trace	2.862	57.245[a]	2.000	40.000	.000	.741
	Roy's Largest Root	2.862	57.245[a]	2.000	40.000	.000	.741

a. Exact statistic

b.

Design: Intercept
Within Subjects Design: COND+LIMB+COND*LIMB

Mauchly's Test of Sphericity[b]

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon[a] Greenhouse-Geisser	Huynh-Feldt	Lower-bound
COND	.873	5.411	2	.067	.888	.925	.500
LIMB	1.000	.000	0	.	1.000	1.000	1.000
COND * LIMB	.719	13.175	2	.001	.781	.806	.500

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.

b.

Design: Intercept
Within Subjects Design: COND+LIMB+COND*LIMB

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
COND	Sphericity Assumed	1914.240	2	957.120	37.530	.000	.478
	Greenhouse-Geisser	1914.240	1.775	1078.226	37.530	.000	.478
	Huynh-Feldt	1914.240	1.850	1034.731	37.530	.000	.478
	Lower-bound	1914.240	1.000	1914.240	37.530	.000	.478
Error(COND)	Sphericity Assumed	2091.236	82	25.503			
	Greenhouse-Geisser	2091.236	72.790	28.730			
	Huynh-Feldt	2091.236	75.850	27.571			
	Lower-bound	2091.236	41.000	51.006			
LIMB	Sphericity Assumed	116.210	1	116.210	.903	.347	.022
	Greenhouse-Geisser	116.210	1.000	116.210	.903	.347	.022
	Huynh-Feldt	116.210	1.000	116.210	.903	.347	.022
	Lower-bound	116.210	1.000	116.210	.903	.347	.022
Error(LIMB)	Sphericity Assumed	5274.144	41	128.638			
	Greenhouse-Geisser	5274.144	41.000	128.638			
	Huynh-Feldt	5274.144	41.000	128.638			
	Lower-bound	5274.144	41.000	128.638			
COND * LIMB	Sphericity Assumed	2083.056	2	1041.528	77.733	.000	.655
	Greenhouse-Geisser	2083.056	1.562	1333.801	77.733	.000	.655
	Huynh-Feldt	2083.056	1.612	1291.837	77.733	.000	.655
	Lower-bound	2083.056	1.000	2083.056	77.733	.000	.655
Error(COND*LIMB)	Sphericity Assumed	1098.705	82	13.399			
	Greenhouse-Geisser	1098.705	64.031	17.159			
	Huynh-Feldt	1098.705	66.112	16.619			
	Lower-bound	1098.705	41.000	26.798			

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	COND	LIMB	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
COND	Linear		642.405	1	642.405	35.389	.000	.463
	Quadratic		1271.836	1	1271.836	38.713	.000	.486
Error(COND)	Linear		744.254	41	18.153			
	Quadratic		1346.983	41	32.853			
LIMB		Linear	116.210	1	116.210	.903	.347	.022
Error(LIMB)		Linear	5274.144	41	128.638			
COND * LIMB	Linear	Linear	420.053	1	420.053	61.193	.000	.599
	Quadratic	Linear	1663.003	1	1663.003	83.429	.000	.670
Error(COND*LIMB)	Linear	Linear	281.441	41	6.864			
	Quadratic	Linear	817.264	41	19.933			

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Intercept	2480417.487	1	2480417.487	11105.277	.000	.996
Error	9157.549	41	223.355			

Appendix E:

Ankle Kinematics General Linear Model: Repeated Measures (Condition x Limb)

Within-Subjects Factors

Measure: MEASURE_1

COND	LIMB	Dependent Variable
1	1	C1LA
	2	C1RA
2	1	C2LA
	2	C2RA
3	1	C3LA
	2	C3RA

Descriptive Statistics

	Mean	Std. Deviation	N
C1LA	22.5001	5.65203	42
C1RA	22.7370	5.67543	42
C2LA	21.3306	5.78938	42
C2RA	17.4400	5.57939	42
C3LA	20.6643	5.20062	42
C3RA	16.1757	3.38528	42

Multivariate Tests[b]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
COND	Pillai's Trace	.698	46.199[a]	2.000	40.000	.000	.698
	Wilks' Lambda	.302	46.199[a]	2.000	40.000	.000	.698
	Hotelling's Trace	2.310	46.199[a]	2.000	40.000	.000	.698
	Roy's Largest Root	2.310	46.199[a]	2.000	40.000	.000	.698
LIMB	Pillai's Trace	.022	.903[a]	1.000	41.000	.347	.022
	Wilks' Lambda	.978	.903[a]	1.000	41.000	.347	.022
	Hotelling's Trace	.022	.903[a]	1.000	41.000	.347	.022
	Roy's Largest Root	.022	.903[a]	1.000	41.000	.347	.022
COND * LIMB	Pillai's Trace	.741	57.245[a]	2.000	40.000	.000	.741
	Wilks' Lambda	.259	57.245[a]	2.000	40.000	.000	.741
	Hotelling's Trace	2.862	57.245[a]	2.000	40.000	.000	.741
	Roy's Largest Root	2.862	57.245[a]	2.000	40.000	.000	.741

a. Exact statistic

b.

Design: Intercept
Within Subjects Design: COND+LIMB+COND*LIMB

Multivariate Tests[b]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
COND	Pillai's Trace	.748	59.385[a]	2.000	40.000	.000	.748
	Wilks' Lambda	.252	59.385[a]	2.000	40.000	.000	.748
	Hotelling's Trace	2.969	59.385[a]	2.000	40.000	.000	.748
	Roy's Largest Root	2.969	59.385[a]	2.000	40.000	.000	.748
LIMB	Pillai's Trace	.390	26.169[a]	1.000	41.000	.000	.390
	Wilks' Lambda	.610	26.169[a]	1.000	41.000	.000	.390
	Hotelling's Trace	.638	26.169[a]	1.000	41.000	.000	.390
	Roy's Largest Root	.638	26.169[a]	1.000	41.000	.000	.390
COND * LIMB	Pillai's Trace	.651	37.282[a]	2.000	40.000	.000	.651
	Wilks' Lambda	.349	37.282[a]	2.000	40.000	.000	.651
	Hotelling's Trace	1.864	37.282[a]	2.000	40.000	.000	.651
	Roy's Largest Root	1.864	37.282[a]	2.000	40.000	.000	.651

a. Exact statistic

b.

Design: Intercept
Within Subjects Design: COND+LIMB+COND*LIMB

Mauchly's Test of Sphericity[b]

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon[a] Greenhouse-Geisser	Huynh-Feldt	Lower-bound
COND	.980	.816	2	.665	.980	1.000	.500
LIMB	1.000	.000	0	.	1.000	1.000	1.000
COND * LIMB	.969	1.255	2	.534	.970	1.000	.500

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.

b.
Design: Intercept
Within Subjects Design: COND+LIMB+COND*LIMB

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
COND	Sphericity Assumed	812.374	2	406.187	53.513	.000	.566
	Greenhouse-Geisser	812.374	1.960	414.391	53.513	.000	.566
	Huynh-Feldt	812.374	2.000	406.187	53.513	.000	.566
	Lower-bound	812.374	1.000	812.374	53.513	.000	.566
Error(COND)	Sphericity Assumed	622.420	82	7.590			
	Greenhouse-Geisser	622.420	80.377	7.744			
	Huynh-Feldt	622.420	82.000	7.590			
	Lower-bound	622.420	41.000	15.181			
LIMB	Sphericity Assumed	464.079	1	464.079	26.169	.000	.390
	Greenhouse-Geisser	464.079	1.000	464.079	26.169	.000	.390
	Huynh-Feldt	464.079	1.000	464.079	26.169	.000	.390
	Lower-bound	464.079	1.000	464.079	26.169	.000	.390
Error(LIMB)	Sphericity Assumed	727.081	41	17.734			
	Greenhouse-Geisser	727.081	41.000	17.734			
	Huynh-Feldt	727.081	41.000	17.734			
	Lower-bound	727.081	41.000	17.734			
COND * LIMB	Sphericity Assumed	278.073	2	139.036	35.149	.000	.462
	Greenhouse-Geisser	278.073	1.940	143.330	35.149	.000	.462
	Huynh-Feldt	278.073	2.000	139.036	35.149	.000	.462
	Lower-bound	278.073	1.000	278.073	35.149	.000	.462
Error(COND*LIMB)	Sphericity Assumed	324.359	82	3.956			
	Greenhouse-Geisser	324.359	79.544	4.078			
	Huynh-Feldt	324.359	82.000	3.956			
	Lower-bound	324.359	41.000	7.911			

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	COND	LIMB	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
COND	Linear		740.365	1	740.365	102.280	.000	.714
	Quadratic		72.010	1	72.010	9.066	.004	.181
Error(COND)	Linear		296.782	41	7.239			
	Quadratic		325.638	41	7.942			
LIMB		Linear	464.079	1	464.079	26.169	.000	.390
Error(LIMB)		Linear	727.081	41	17.734			
COND * LIMB	Linear	Linear	234.467	1	234.467	70.440	.000	.632
	Quadratic	Linear	43.606	1	43.606	9.516	.004	.188
Error(COND*LIMB)	Linear	Linear	136.473	41	3.329			
	Quadratic	Linear	187.886	41	4.583			

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Intercept	102229.244	1	102229.244	808.245	.000	.952
Error	5185.806	41	126.483			

Appendix F:

Tukeys Post-Hoc calculations for contralateral (non braced) and ipsilateral (braced) hip flexion across conditions

Contralateral Hip Mean ± SD	Normal Hip Flexion:	Knee Restricted Hip Flexion:	Ankle Restricted Hip Flexion:
Non braced Hip Flexion: 103.7°±13.2°	______	-0.6°	-2.3°*
Knee Restricted Hip Flexion: 103.1°±12.6°		______	-1.7°
Ankle Restricted Hip Flexion: 101.4°±13.2°			______

*p< .05; $_{.95}q_{65,6}$≈ 4.16; MSe=11.7 (Huynh-Feidt correction); N=42; 2.2° difference needed for significance.

Ipsilateral Hip Mean ± SD	Normal Hip Flexion:	Knee Restricted Hip Flexion:	Ankle Restricted Hip Flexion:
Non braced Hip Flexion: 113.2°±11.9°	______	-2.4°*	-0.6°
Knee Restricted Hip Flexion: 110.8°±13.2°		______	+1.8°
Ankle Restricted Hip Flexion: 112.7°±13.4°			______

*p< .05; $_{.95}q_{65,6}$≈ 4.16; MSe=4.2 (Huynh-Feidt correction); N=42; 2.2° difference needed for significance.

Appendix G:

Tukeys Post-Hoc calculations for contralateral (non braced) and ipsilateral (braced) knee flexion joint displacement across conditions

Contralateral Knee Mean ± SD	Normal Knee Flexion:	Knee Restricted Knee Flexion:	Ankle Restricted Knee Flexion:
Non braced Knee Flexion: 98.7°±8.9°	______	+0.3°	-0.7°
Knee Restricted Knee Flexion: 98.9°±8.9°		______	-1.1°
Ankle Restricted Knee Flexion: 97.9°±8.1°			______

*p< .05; ${}_{.95}q_{65,6}\approx 4.16$; MSe=17.2 (Huynh-Feidt correction); N=42; 2.7° difference needed for significance.

Ipsilateral Knee Mean ± SD	Normal Knee Flexion:	Knee Restricted Knee Flexion:	Ankle Restricted Knee Flexion:
Non braced Knee Flexion: 106.8°±8.8°	______	-13.8°*	-7.1°*
Knee Restricted Knee Flexion: 93.8°±6.8°		______	+6.7°*
Ankle Restricted Knee Flexion: 99.8°±9.0°			______

*p< .05; ${}_{.95}q_{65,6}\approx 4.16$; MSe=17.2 (Huynh-Feidt correction); N=42; 2.7° difference needed for significance.

Appendix H:

Tukeys Post-Hoc calculations for contralateral (non braced) and ipsilateral (braced) ankle joint dorsiflexion displacement across conditions

Contralateral Ankle Mean ± SD	Normal Ankle Dorsiflexion:	Knee Restricted Ankle Dorsiflexion:	Ankle Restricted Ankle Dorsiflexion:
Non braced Ankle Dorsiflexion:22.5°±5.7°	______	-1.2°	-1.8°*
Knee Restricted Ankle Dorsiflexion: 21.3°±5.8°		______	-0.7°
Ankle Restricted Ankle Dorsiflexion: 20.7°±5.2°			______

*p< .05; $_{.95}q_{65,6} \approx 4.16$; MSe=4.0 (Huynh-Feidt correction); N=42; 1.3° difference needed for significance.

Ipsilateral Ankle Mean ± SD	Normal Ankle Dorsiflexion:	Knee Restricted Ankle Dorsiflexion:	Ankle Restricted Ankle Dorsiflexion:
Non braced Ankle Dorsiflexion:22.7°±5.7°	______	-5.3°*	-6.6°*
Knee Restricted Ankle Dorsiflexion: 17.4°±5.6°		______	-1.3°*
Ankle Restricted Ankle Dorsiflexion: 16.2°±3.4°			______

*p< .05; $_{.95}q_{65,6} \approx 4.16$; MSe=4.0 (Huynh-Feidt correction); N=42; 1.3° difference needed for significance.

Appendix I:

Energetic General Linear Model: Repeated Measures (Condition x Limb x Joint)

Within-Subjects Factors

Measure: MEASURE_1

COND	LIMB	JOINT	Dependent Variable
1	1	1	LHC1
		2	LKC1
		3	LAC1
	2	1	RHC1
		2	RKC1
		3	RAC1
2	1	1	LHC2
		2	LKC2
		3	LAC2
	2	1	RHC2
		2	RKC2
		3	RAC2
3	1	1	LHC3
		2	LKC3
		3	LAC3
	2	1	RHC3
		2	RKC3
		3	RAC3

Descriptive Statistics

	Mean	Std. Deviation	N
Left C1	.1308	.08465	42
LKC1	.4265	.12497	42
LAC1	.0149	.01596	42
RHC1	.2124	.18932	42
RKC1	.4906	.23377	42
RAC1	.0441	.05132	42
LHC2	.1386	.11359	42
LKC2	.4531	.13765	42
LAC2	.0124	.01046	42
RHC2	.2168	.18318	42
RKC2	.4040	.19714	42
RAC2	.0647	.04953	42
LHC3	.1097	.08460	42
LKC3	.4576	.13245	42
LAC3	.0111	.01166	42
RHC3	.1995	.19805	42
RKC3	.4374	.21362	42
RAC3	.0323	.03804	42

Multivariate Tests[b]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
COND	Pillai's Trace	.137	3.179[a]	2.000	40.000	.052	.137
	Wilks' Lambda	.863	3.179[a]	2.000	40.000	.052	.137
	Hotelling's Trace	.159	3.179[a]	2.000	40.000	.052	.137
	Roy's Largest Root	.159	3.179[a]	2.000	40.000	.052	.137
LIMB	Pillai's Trace	.283	16.213[a]	1.000	41.000	.000	.283
	Wilks' Lambda	.717	16.213[a]	1.000	41.000	.000	.283
	Hotelling's Trace	.395	16.213[a]	1.000	41.000	.000	.283
	Roy's Largest Root	.395	16.213[a]	1.000	41.000	.000	.283
JOINT	Pillai's Trace	.920	230.766[a]	2.000	40.000	.000	.920
	Wilks' Lambda	.080	230.766[a]	2.000	40.000	.000	.920
	Hotelling's Trace	11.538	230.766[a]	2.000	40.000	.000	.920
	Roy's Largest Root	11.538	230.766[a]	2.000	40.000	.000	.920
COND * LIMB	Pillai's Trace	.239	6.268[a]	2.000	40.000	.004	.239
	Wilks' Lambda	.761	6.268[a]	2.000	40.000	.004	.239
	Hotelling's Trace	.313	6.268[a]	2.000	40.000	.004	.239
	Roy's Largest Root	.313	6.268[a]	2.000	40.000	.004	.239
COND * JOINT	Pillai's Trace	.378	5.766[a]	4.000	38.000	.001	.378
	Wilks' Lambda	.622	5.766[a]	4.000	38.000	.001	.378
	Hotelling's Trace	.607	5.766[a]	4.000	38.000	.001	.378
	Roy's Largest Root	.607	5.766[a]	4.000	38.000	.001	.378
LIMB * JOINT	Pillai's Trace	.076	1.635[a]	2.000	40.000	.208	.076
	Wilks' Lambda	.924	1.635[a]	2.000	40.000	.208	.076
	Hotelling's Trace	.082	1.635[a]	2.000	40.000	.208	.076
	Roy's Largest Root	.082	1.635[a]	2.000	40.000	.208	.076
COND * LIMB * JOINT	Pillai's Trace	.532	10.786[a]	4.000	38.000	.000	.532
	Wilks' Lambda	.468	10.786[a]	4.000	38.000	.000	.532
	Hotelling's Trace	1.135	10.786[a]	4.000	38.000	.000	.532
	Roy's Largest Root	1.135	10.786[a]	4.000	38.000	.000	.532

a. Exact statistic

b.
Design: Intercept
Within Subjects Design: COND+LIMB+JOINT+COND*LIMB+COND*JOINT+LIMB*JOINT+COND*LIMB*JOINT

Mauchly's Test of Sphericity[b]

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon[a] Greenhouse-Geisser	Huynh-Feldt	Lower-bound
COND	.997	.112	2	.945	.997	1.000	.500
LIMB	1.000	.000	0	.	1.000	1.000	1.000
JOINT	.563	22.954	2	.000	.696	.713	.500
COND * LIMB	.799	8.986	2	.011	.832	.863	.500
COND * JOINT	.025	145.447	9	.000	.474	.496	.250
LIMB * JOINT	.518	26.274	2	.000	.675	.690	.500
COND * LIMB * JOINT	.244	55.610	9	.000	.616	.658	.250

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.

b.
Design: Intercept
Within Subjects Design: COND+LIMB+JOINT+COND*LIMB+COND*JOINT+LIMB*JOINT+COND*LIMB*JOINT

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
COND	Sphericity Assumed	.018	2	.009	3.088	.051	.070
	Greenhouse-Geisser	.018	1.994	.009	3.088	.051	.070
	Huynh-Feldt	.018	2.000	.009	3.088	.051	.070
	Lower-bound	.018	1.000	.018	3.088	.086	.070
Error(COND)	Sphericity Assumed	.241	82	.003			
	Greenhouse-Geisser	.241	81.771	.003			
	Huynh-Feldt	.241	82.000	.003			
	Lower-bound	.241	41.000	.006			
LIMB	Sphericity Assumed	.281	1	.281	16.213	.000	.283
	Greenhouse-Geisser	.281	1.000	.281	16.213	.000	.283
	Huynh-Feldt	.281	1.000	.281	16.213	.000	.283
	Lower-bound	.281	1.000	.281	16.213	.000	.283
Error(LIMB)	Sphericity Assumed	.710	41	.017			
	Greenhouse-Geisser	.710	41.000	.017			
	Huynh-Feldt	.710	41.000	.017			
	Lower-bound	.710	41.000	.017			
JOINT	Sphericity Assumed	22.505	2	11.253	188.893	.000	.822
	Greenhouse-Geisser	22.505	1.392	16.166	188.893	.000	.822
	Huynh-Feldt	22.505	1.426	15.785	188.893	.000	.822
	Lower-bound	22.505	1.000	22.505	188.893	.000	.822
Error(JOINT)	Sphericity Assumed	4.885	82	.060			
	Greenhouse-Geisser	4.885	57.078	.086			
	Huynh-Feldt	4.885	58.454	.084			
	Lower-bound	4.885	41.000	.119			
COND * LIMB	Sphericity Assumed	.037	2	.019	7.071	.001	.147
	Greenhouse-Geisser	.037	1.665	.022	7.071	.003	.147
	Huynh-Feldt	.037	1.727	.022	7.071	.003	.147
	Lower-bound	.037	1.000	.037	7.071	.011	.147
Error(COND*LIMB)	Sphericity Assumed	.216	82	.003			
	Greenhouse-Geisser	.216	68.265	.003			
	Huynh-Feldt	.216	70.805	.003			
	Lower-bound	.216	41.000	.005			
COND * JOINT	Sphericity Assumed	.057	4	.014	2.267	.064	.052
	Greenhouse-Geisser	.057	1.895	.030	2.267	.113	.052
	Huynh-Feldt	.057	1.984	.028	2.267	.111	.052
	Lower-bound	.057	1.000	.057	2.267	.140	.052
Error(COND*JOINT)	Sphericity Assumed	1.022	164	.006			
	Greenhouse-Geisser	1.022	77.685	.013			
	Huynh-Feldt	1.022	81.338	.013			
	Lower-bound	1.022	41.000	.025			
LIMB * JOINT	Sphericity Assumed	.229	2	.114	2.674	.075	.061
	Greenhouse-Geisser	.229	1.350	.169	2.674	.097	.061
	Huynh-Feldt	.229	1.380	.166	2.674	.096	.061
	Lower-bound	.229	1.000	.229	2.674	.110	.061
Error(LIMB*JOINT)	Sphericity Assumed	3.507	82	.043			
	Greenhouse-Geisser	3.507	55.349	.063			
	Huynh-Feldt	3.507	56.561	.062			
	Lower-bound	3.507	41.000	.086			
COND * LIMB * JOINT	Sphericity Assumed	.121	4	.030	7.203	.000	.149
	Greenhouse-Geisser	.121	2.463	.049	7.203	.001	.149
	Huynh-Feldt	.121	2.633	.046	7.203	.000	.149
	Lower-bound	.121	1.000	.121	7.203	.010	.149
Error(COND*LIMB*JOINT)	Sphericity Assumed	.688	164	.004			
	Greenhouse-Geisser	.688	100.991	.007			
	Huynh-Feldt	.688	107.939	.006			
	Lower-bound	.688	41.000	.017			

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	COND	LIMB	JOINT	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
COND	Linear			.018	1	.018	6.434	.015	.136
	Quadratic			.000	1	.000	.057	.812	.001
Error(COND)	Linear			.115	41	.003			
	Quadratic			.126	41	.003			
LIMB		Linear		.281	1	.281	16.213	.000	.283
Error(LIMB)		Linear		.710	41	.017			
JOINT			Linear	2.401	1	2.401	99.974	.000	.709
			Quadratic	20.104	1	20.104	211.338	.000	.838
Error(JOINT)			Linear	.985	41	.024			
			Quadratic	3.900	41	.095			
COND * LIMB	Linear	Linear		.025	1	.025	12.463	.001	.233
	Quadratic	Linear		.012	1	.012	3.785	.059	.085
Error(COND*LIMB)	Linear	Linear		.082	41	.002			
	Quadratic	Linear		.134	41	.003			
COND * JOINT	Linear		Linear	.002	1	.002	.700	.408	.017
			Quadratic	5.282E-05	1	5.282E-05	.011	.919	.000
	Quadratic		Linear	7.621E-05	1	7.621E-05	.014	.907	.000
			Quadratic	.055	1	.055	4.603	.038	.101
Error(COND*JOINT)	Linear		Linear	.103	41	.003			
			Quadratic	.205	41	.005			
	Quadratic		Linear	.227	41	.006			
			Quadratic	.487	41	.012			
LIMB * JOINT		Linear	Linear	.075	1	.075	2.849	.099	.065
			Quadratic	.153	1	.153	2.596	.115	.060
Error(LIMB*JOINT)		Linear	Linear	1.086	41	.026			
			Quadratic	2.421	41	.059			
COND * LIMB * JOINT	Linear	Linear	Linear	.001	1	.001	.490	.488	.012
			Quadratic	.050	1	.050	18.846	.000	.315
	Quadratic	Linear	Linear	.008	1	.008	2.002	.165	.047
			Quadratic	.061	1	.061	8.597	.005	.173
Error(COND*LIMB*JOINT)	Linear	Linear	Linear	.116	41	.003			
			Quadratic	.109	41	.003			
	Quadratic	Linear	Linear	.171	41	.004			
			Quadratic	.292	41	.007			

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Intercept	34.707	1	34.707	585.370	.000	.935
Error	2.431	41	.059			

Appendix J:

Energetic General Linear Model: Repeated Measures (Condition x Limb): Hip

Within-Subjects Factors

Measure: MEASURE_1

COND	LIMB	Dependent Variable
1	1	LHC1
	2	RHC1
2	1	LHC2
	2	RHC2
3	1	LHC3
	2	RHC3

Descriptive Statistics

	Mean	Std. Deviation	N
Left C1	.1308	.08465	42
RHC1	.2124	.18932	42
LHC2	.1386	.11359	42
RHC2	.2168	.18318	42
LHC3	.1097	.08460	42
RHC3	.1995	.19805	42

Multivariate Tests[b]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
COND	Pillai's Trace	.064	1.360[a]	2.000	40.000	.268	.064
	Wilks' Lambda	.936	1.360[a]	2.000	40.000	.268	.064
	Hotelling's Trace	.068	1.360[a]	2.000	40.000	.268	.064
	Roy's Largest Root	.068	1.360[a]	2.000	40.000	.268	.064
LIMB	Pillai's Trace	.175	8.700[a]	1.000	41.000	.005	.175
	Wilks' Lambda	.825	8.700[a]	1.000	41.000	.005	.175
	Hotelling's Trace	.212	8.700[a]	1.000	41.000	.005	.175
	Roy's Largest Root	.212	8.700[a]	1.000	41.000	.005	.175
COND * LIMB	Pillai's Trace	.008	.170[a]	2.000	40.000	.844	.008
	Wilks' Lambda	.992	.170[a]	2.000	40.000	.844	.008
	Hotelling's Trace	.008	.170[a]	2.000	40.000	.844	.008
	Roy's Largest Root	.008	.170[a]	2.000	40.000	.844	.008

a. Exact statistic

b.

Design: Intercept
Within Subjects Design: COND+LIMB+COND*LIMB

Mauchly's Test of Sphericity[b]

Measure: MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilon[a] Greenhouse-Geisser	Huynh-Feldt	Lower-bound
COND	.906	3.948	2	.139	.914	.955	.500
LIMB	1.000	.000	0	.	1.000	1.000	1.000
COND * LIMB	.845	6.726	2	.035	.866	.901	.500

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.

b.

Design: Intercept
Within Subjects Design: COND+LIMB+COND*LIMB

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
COND	Sphericity Assumed	.024	2	.012	1.239	.295	.029
	Greenhouse-Geisser	.024	1.828	.013	1.239	.293	.029
	Huynh-Feldt	.024	1.909	.013	1.239	.294	.029
	Lower-bound	.024	1.000	.024	1.239	.272	.029
Error(COND)	Sphericity Assumed	.795	82	.010			
	Greenhouse-Geisser	.795	74.956	.011			
	Huynh-Feldt	.795	78.274	.010			
	Lower-bound	.795	41.000	.019			
LIMB	Sphericity Assumed	.436	1	.436	8.700	.005	.175
	Greenhouse-Geisser	.436	1.000	.436	8.700	.005	.175
	Huynh-Feldt	.436	1.000	.436	8.700	.005	.175
	Lower-bound	.436	1.000	.436	8.700	.005	.175
Error(LIMB)	Sphericity Assumed	2.053	41	.050			
	Greenhouse-Geisser	2.053	41.000	.050			
	Huynh-Feldt	2.053	41.000	.050			
	Lower-bound	2.053	41.000	.050			
COND * LIMB	Sphericity Assumed	.001	2	.001	.113	.893	.003
	Greenhouse-Geisser	.001	1.732	.001	.113	.866	.003
	Huynh-Feldt	.001	1.802	.001	.113	.874	.003
	Lower-bound	.001	1.000	.001	.113	.738	.003
Error(COND*LIMB)	Sphericity Assumed	.539	82	.007			
	Greenhouse-Geisser	.539	71.010	.008			
	Huynh-Feldt	.539	73.862	.007			
	Lower-bound	.539	41.000	.013			

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	COND	LIMB	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
COND	Linear		.012	1	.012	1.798	.187	.042
	Quadratic		.012	1	.012	.942	.337	.022
Error(COND)	Linear		.276	41	.007			
	Quadratic		.519	41	.013			
LIMB		Linear	.436	1	.436	8.700	.005	.175
Error(LIMB)		Linear	2.053	41	.050			
COND * LIMB	Linear	Linear	.001	1	.001	.141	.709	.003
	Quadratic	Linear	.001	1	.001	.096	.758	.002
Error(COND*LIMB)	Linear	Linear	.205	41	.005			
	Quadratic	Linear	.333	41	.008			

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Intercept	7.110	1	7.110	133.582	.000	.765
Error	2.182	41	.053			

Appendix K:

Energetic General Linear Model: Repeated Measures (Condition x Limb): Knee

Within-Subjects Factors

Measure: MEASURE_1

COND	LIMB	Dependent Variable
1	1	LKC1
	2	RKC1
2	1	LKC2
	2	RKC2
3	1	LKC3
	2	RKC3

Descriptive Statistics

	Mean	Std. Deviation	N
LKC1	.4265	.12497	42
RKC1	.4906	.23377	42
LKC2	.4531	.13765	42
RKC2	.4040	.19714	42
LKC3	.4576	.13245	42
RKC3	.4374	.21362	42

Multivariate Tests[b]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
COND	Pillai's Trace	.141	3.279[a]	2.000	40.000	.048	.141
	Wilks' Lambda	.859	3.279[a]	2.000	40.000	.048	.141
	Hotelling's Trace	.164	3.279[a]	2.000	40.000	.048	.141
	Roy's Largest Root	.164	3.279[a]	2.000	40.000	.048	.141
LIMB	Pillai's Trace	.000	.004[a]	1.000	41.000	.952	.000
	Wilks' Lambda	1.000	.004[a]	1.000	41.000	.952	.000
	Hotelling's Trace	.000	.004[a]	1.000	41.000	.952	.000
	Roy's Largest Root	.000	.004[a]	1.000	41.000	.952	.000
COND * LIMB	Pillai's Trace	.498	19.851[a]	2.000	40.000	.000	.498
	Wilks' Lambda	.502	19.851[a]	2.000	40.000	.000	.498
	Hotelling's Trace	.993	19.851[a]	2.000	40.000	.000	.498
	Roy's Largest Root	.993	19.851[a]	2.000	40.000	.000	.498

a. Exact statistic

b.
Design: Intercept
Within Subjects Design: COND+LIMB+COND*LIMB

Mauchly's Test of Sphericity[b]

Measure: MEASURE_1

					Epsilon[a]		
Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Greenhouse-Geisser	Huynh-Feldt	Lower-bound
COND	.806	8.651	2	.013	.837	.869	.500
LIMB	1.000	.000	0	.	1.000	1.000	1.000
COND * LIMB	.740	12.051	2	.002	.794	.820	.500

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.

b.
Design: Intercept
Within Subjects Design: COND+LIMB+COND*LIMB

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
COND	Sphericity Assumed	.039	2	.019	3.806	.026	.085
	Greenhouse-Geisser	.039	1.674	.023	3.806	.034	.085
	Huynh-Feldt	.039	1.737	.022	3.806	.032	.085
	Lower-bound	.039	1.000	.039	3.806	.058	.085
Error(COND)	Sphericity Assumed	.416	82	.005			
	Greenhouse-Geisser	.416	68.648	.006			
	Huynh-Feldt	.416	71.232	.006			
	Lower-bound	.416	41.000	.010			
LIMB	Sphericity Assumed	.000	1	.000	.004	.952	.000
	Greenhouse-Geisser	.000	1.000	.000	.004	.952	.000
	Huynh-Feldt	.000	1.000	.000	.004	.952	.000
	Lower-bound	.000	1.000	.000	.004	.952	.000
Error(LIMB)	Sphericity Assumed	2.099	41	.051			
	Greenhouse-Geisser	2.099	41.000	.051			
	Huynh-Feldt	2.099	41.000	.051			
	Lower-bound	2.099	41.000	.051			
COND * LIMB	Sphericity Assumed	.146	2	.073	17.527	.000	.299
	Greenhouse-Geisser	.146	1.587	.092	17.527	.000	.299
	Huynh-Feldt	.146	1.641	.089	17.527	.000	.299
	Lower-bound	.146	1.000	.146	17.527	.000	.299
Error(COND*LIMB)	Sphericity Assumed	.341	82	.004			
	Greenhouse-Geisser	.341	65.073	.005			
	Huynh-Feldt	.341	67.264	.005			
	Lower-bound	.341	41.000	.008			

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	COND	LIMB	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
COND	Linear		.005	1	.005	1.790	.188	.042
	Quadratic		.034	1	.034	4.593	.038	.101
Error(COND)	Linear		.117	41	.003			
	Quadratic		.299	41	.007			
LIMB		Linear	.000	1	.000	.004	.952	.000
Error(LIMB)		Linear	2.099	41	.051			
COND * LIMB	Linear	Linear	.075	1	.075	35.029	.000	.461
	Quadratic	Linear	.071	1	.071	11.475	.002	.219
Error(COND*LIMB)	Linear	Linear	.087	41	.002			
	Quadratic	Linear	.253	41	.006			

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Intercept	49.876	1	49.876	410.210	.000	.909
Error	4.985	41	.122			

Appendix L:

Energetic General Linear Model: Repeated Measures (Condition x Limb): Ankle

Within-Subjects Factors

Measure: MEASURE_1

COND	LIMB	Dependent Variable
1	1	LAC1
	2	RAC1
2	1	LAC2
	2	RAC2
3	1	LAC3
	2	RAC3

Descriptive Statistics

	Mean	Std. Deviation	N
LAC1	.0149	.01596	42
RAC1	.0441	.05132	42
LAC2	.0124	.01046	42
RAC2	.0647	.04953	42
LAC3	.0111	.01166	42
RAC3	.0323	.03804	42

Multivariate Tests[b]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared
COND	Pillai's Trace	.308	8.910[a]	2.000	40.000	.001	.308
	Wilks' Lambda	.692	8.910[a]	2.000	40.000	.001	.308
	Hotelling's Trace	.445	8.910[a]	2.000	40.000	.001	.308
	Roy's Largest Root	.445	8.910[a]	2.000	40.000	.001	.308
LIMB	Pillai's Trace	.533	46.765[a]	1.000	41.000	.000	.533
	Wilks' Lambda	.467	46.765[a]	1.000	41.000	.000	.533
	Hotelling's Trace	1.141	46.765[a]	1.000	41.000	.000	.533
	Roy's Largest Root	1.141	46.765[a]	1.000	41.000	.000	.533
COND * LIMB	Pillai's Trace	.497	19.756[a]	2.000	40.000	.000	.497
	Wilks' Lambda	.503	19.756[a]	2.000	40.000	.000	.497
	Hotelling's Trace	.988	19.756[a]	2.000	40.000	.000	.497
	Roy's Largest Root	.988	19.756[a]	2.000	40.000	.000	.497

a. Exact statistic

b.

Design: Intercept
Within Subjects Design: COND+LIMB+COND*LIMB

Mauchly's Test of Sphericity[b]

Measure: MEASURE_1

					Epsilon[a]		
Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Greenhouse-Geisser	Huynh-Feldt	Lower-bound
COND	.870	5.574	2	.062	.885	.922	.500
LIMB	1.000	.000	0	.	1.000	1.000	1.000
COND * LIMB	.895	4.420	2	.110	.905	.945	.500

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.

b.

Design: Intercept
Within Subjects Design: COND+LIMB+COND*LIMB

Tests of Within-Subjects Effects

Measure: MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
COND	Sphericity Assumed	.012	2	.006	9.623	.000	.190
	Greenhouse-Geisser	.012	1.770	.007	9.623	.000	.190
	Huynh-Feldt	.012	1.844	.006	9.623	.000	.190
	Lower-bound	.012	1.000	.012	9.623	.003	.190
Error(COND)	Sphericity Assumed	.051	82	.001			
	Greenhouse-Geisser	.051	72.561	.001			
	Huynh-Feldt	.051	75.594	.001			
	Lower-bound	.051	41.000	.001			
LIMB	Sphericity Assumed	.074	1	.074	46.765	.000	.533
	Greenhouse-Geisser	.074	1.000	.074	46.765	.000	.533
	Huynh-Feldt	.074	1.000	.074	46.765	.000	.533
	Lower-bound	.074	1.000	.074	46.765	.000	.533
Error(LIMB)	Sphericity Assumed	.065	41	.002			
	Greenhouse-Geisser	.065	41.000	.002			
	Huynh-Feldt	.065	41.000	.002			
	Lower-bound	.065	41.000	.002			
COND * LIMB	Sphericity Assumed	.011	2	.005	18.521	.000	.311
	Greenhouse-Geisser	.011	1.811	.006	18.521	.000	.311
	Huynh-Feldt	.011	1.889	.006	18.521	.000	.311
	Lower-bound	.011	1.000	.011	18.521	.000	.311
Error(COND*LIMB)	Sphericity Assumed	.024	82	.000			
	Greenhouse-Geisser	.024	74.235	.000			
	Huynh-Feldt	.024	77.466	.000			
	Lower-bound	.024	41.000	.001			

Tests of Within-Subjects Contrasts

Measure: MEASURE_1

Source	COND	LIMB	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
COND	Linear		.003	1	.003	3.571	.066	.080
	Quadratic		.009	1	.009	17.984	.000	.305
Error(COND)	Linear		.030	41	.001			
	Quadratic		.021	41	.001			
LIMB		Linear	.074	1	.074	46.765	.000	.533
Error(LIMB)		Linear	.065	41	.002			
COND * LIMB	Linear	Linear	.001	1	.001	2.027	.162	.047
	Quadratic	Linear	.010	1	.010	40.277	.000	.496
Error(COND*LIMB)	Linear	Linear	.014	41	.000			
	Quadratic	Linear	.010	41	.000			

Tests of Between-Subjects Effects

Measure: MEASURE_1
Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared
Intercept	.226	1	.226	62.365	.000	.603
Error	.148	41	.004			

Appendix M:

Tukeys Post-Hoc calculations for contralateral (non braced) and ipsilateral (braced) hip energetics across conditions

Contralateral (Left) Hip Energetics 1)Non braced, 2)Knee Restricted, 3)Ankle Restricted	Contralateral Mean Hip Difference
Non braced Hip Work (1) Mean: Contralateral 0.13±.08 Nm/kg	(2)-0.01 (3)+0.02
Knee Restricted Hip Work (2) Mean: Contralateral 0.14±.11 Nm/kg	(1)+0.01 (3)+0.03
Ankle Restricted Hip Work (3) Mean: Contralateral 0.11±.08 Nm/kg	(1)-0.02 (2)-0.03
*p< .05; ${}_{.95}q_{6,82}\approx 4.163$; MSe=0.007 (Huynh-Feidt correction); N=42; 0.05 Nm/kg difference needed for significance.	

Ipsilateral (Right) Hip Energetics 1)Non braced, 2)Knee Restricted, 3)Ankle Restricted	Ipsilateral Mean Hip Difference
Non braced Hip Work (1) Mean: Ipsilateral 0.21±0.19 Nm/kg	(2)-0.005 (3)+0.01
Knee Restricted Hip Work (2) Mean: Ipsilateral 0.22±0.18 Nm/kg	(1)+0.004 (3)+0.02
Ankle Restricted Hip Work (3) Mean: Ipsilateral 0.20±0.20 Nm/kg	(1)-0.01 (2)-0.02
*p< .05; ${}_{.95}q_{6,82}\approx 4.163$; MSe=0.007 (Huynh-Feidt correction); N=42; 0.05 Nm/kg difference needed for significance.	

Appendix N:

Tukeys Post-Hoc calculations for contralateral (non braced) and ipsilateral (braced) knee energetics across conditions

Contralateral (Left) Knee Energetics 1)Non braced, 2)Knee Restricted, 3)Ankle Restricted	Contralateral Mean Knee Difference
Non braced Knee Work (1) Mean: Contralateral 0.43±0.12 Nm/kg	(2)-0.02 (3)-0.03
Knee Restricted Knee Work (2) Mean: Contralateral 0.45±0.14 Nm/kg	(1)+0.02 (3)-0.004
Ankle Restricted Knee Work (3) Mean: Contralateral 0.45±0.13 Nm/kg	(1)+0.03 (2)+0.004
*p< .05; $_{.95}q_{6,82}$≈ 4.163; MSe=0.004 (Huynh-Feidt correction); N=42; 0.05 Nm/kg difference needed for significance.	

Ipsilateral (Right) Knee Energetics 1)Non braced, 2)Knee Restricted, 3)Ankle Restricted	Ipsilateral Mean Knee Difference
Non braced Knee Work (1) Mean: Ipsilateral 0.49±0.23 Nm/kg	(2)+0.07* (3)+0.04
Knee Restricted Knee Work (2) Mean: Ipsilateral 0.40±0.20 Nm/kg	(1)-0.07* (3)-0.03
Ankle Restricted Knee Work (3) Mean: Ipsilateral 0.44±0.21 Nm/kg	(1)-0.04 (2)+0.03
*p< .05; $_{.95}q_{6,82}$≈ 4.163; MSe=0.004 (Huynh-Feidt correction); N=42; 0.05 Nm/kg difference needed for significance.	

Appendix O:

Tukeys Post-Hoc calculations for contralateral (non braced) and ipsilateral (braced) ankle energetics across conditions

Contralateral (Left) Ankle Energetics 1)Non braced, 2)Knee Restricted, 3)Ankle Restricted	Contralateral Mean Ankle Difference
Noon braced Ankle Work (1)	(2)+0.002
Mean: Contralateral 0.015±0.016 Nm/kg	(3)+0.004
Knee Restricted Ankle Work (2)	(1)-0.003
Mean: Contralateral 0.012±0.012 Nm/kg	(3)+0.001
Ankle Restricted Ankle Work (3)	(1)-0.004
Mean: Contralateral 0.011±0.012 Nm/kg	(2)-0.001
*p< .05; $_{.95}q_{6,82}\approx$ 4.163; MSe=0.0005; N=42; 0.014 Nm/kg difference needed for significance.	

Ipsilateral (Right) Ankle Energetics 1)Non braced, 2)Knee Restricted, 3)Ankle Restricted	Ipsilateral Mean Ankle Difference
Non braced Ankle Work (1)	(2)-0.02*
Mean: Ipsilateral 0.044±0.051 Nm/kg	(3)+0.01*
Knee Restricted Ankle Work (2)	(1)+0.02*
Mean: Ipsilateral 0.065±0.050 Nm/kg	(3) +0.03*
Ankle Restricted Ankle Work (3)	(1)-0.01*
Mean: Ipsilateral 0.032±0.038 Nm/kg	(2)-0.03*
*p< .05; $_{.95}q_{6,82}\approx$ 4.163; MSe=0.0005; N=42; 0.014 Nm/kg difference needed for significance.	

Appendix P: IRB

The University of North Carolina at Greensboro

CONSENT TO ACT AS A HUMAN SUBJECT

Short Form
(an Oral Presentation must be used with this form)

Project Title: Kinematic and Kinetic Effects of Knee and Ankle Sagittal Plane Joint Restricitons During Squatting

Project Director: Lee Howard PT, ATC, CSCS

Subject's Name: ____

Dateof Consent: ____

Lee Howard has explained in the preceding oral presentation the procedures involved in this research project including the purpose and what will be required of you. Any benefits and risks were also described. It is understood that if you have received medical treatment for any knee condition over the last 3 months that you are excluded from this study. Lee Howard has answered all of your current questions regarding your participation in this project. You are free to refuse to participate or to withdraw your consent to participate in this research at any time without penalty or prejudice; your participation is entirely voluntary. Your privacy will be protected because you will not be identified by name as a participant in this project.

The research and this consent form have been approved by the University of North Carolina at Greensboro Institutional Review Board, which insures that research involving people follows federal regulations. Questions regarding your rights as a participant in this project can be answered by calling Dr. Beverly Maddox-Britt at (336) 334-5878. Questions regarding the research itself will be answered by Lee Howard by calling 287-5526. Any new information that develops during the project will be provided to you if the information might affect your willingness to continue participation in the project.

By signing this form, you are agreeing to participate in the project described to you by Lee Howard.

____
Subject's Signature

____
Witness to Oral Presentation and Subject's Signature

ORAL PRESENTATION

(must accompany Short Consent Form)

1. Explanation of research purpose and procedures

You are being asked to participate in a study evaluating the difference between unrestricted squats and squats with an induced knee and ankle joint restriction. The knee restriction will be created by a knee brace allowing only 90° of bend. The ankle restriction will be created by a board that will prevent greater than 10° of anterior knee movement referenced from the ankle. In each condition you will squat down until your rear makes slight contact with the bench and then return to the upright position. Data will be collected from eight motion sensors that will be secured to you by tape and/ or velcro. In order to qualify for this investigation, you must be recreationally active (participate in physical activity at least 3 times per week) and have a history of using squats or similar exercises in your training regimen. You may not participate in this study if you have had any reconstructive knee surgery or received medical treatment for knee pain over the last 6 months. If you meet these criteria, you will be asked to attend one 60 minute testing session. At the testing session, you will be asked to perform a series of 3 squats in each of the 3 conditions:

1. Parallel thigh squat with a standardized stance width
2. Squat with an induced knee range of motion restriction (90°) using the same stance width and squat depth parameters as in condition 1.
3. Squat with an induced ankle range of motion restriction (10°) using the same stance width and squat depth parameters as in condition 1.

Each subject will perform the squat standing in front of an adjustable bench to allow a parallel thigh position (approximately 110° of knee bend). Subjects will be instructed to look straight ahead with their arms outstretched to a parallel to floor position using their standardized stances on the force plates. Several practice repetitions will be allowed before the 3 test repetitions in each of the 3 conditions will be recorded. This will serve as a specific warm up.

Prior to the exercises, a total of eight small motion sensors (less than 1"x1"x1") will be placed on your feet, legs, and torso for the purpose of data collection.

2. Benefits

$15 compensation after completion of the trials. No other direct benefits to you as a subject.

3. Risks

There is a slight risk of muscle soreness during participation in the study procedures. Contact Dr. Beverly Maddox-Britt at (336) 334-5878 about any research-related injuries.

4. The opportunity to withdraw without penalty
   You have the opportunity to withdraw from this study at any time without penalty.

5. The opportunity to ask questions
   You may ask questions at any time during the study.

6. The amount of time required of the subjects
   No more than 60 minutes will be required to complete the entire study.

7. Confidentiality of data and final disposition of data
   All the data associated with your visit to the laboratory will be identified with code numbers. Upon completion of the study the principal investigator will store all data.

____

Signature of Person Obtaining Consent on
Behalf of UNCG and Date